BIOMECHANICAL ADAPTATIONS INVOLVED IN RAMP DESCENT: IMPACT OF MICROPROCESSOR-CONTROLLED ANKLE-FOOT PROSTHESIS

Kinetic and kinematic responses to using microprocessor-controlled ankle-foot prosthesis in unilateral trans-tibial amputees during ramp descent

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Abstract

Ramp descent is a demanding task for trans-tibial amputees, due to the difficulty in controlling body weight progression over the prosthetic foot. A deeper understanding of the impact of foot function on ramp descent biomechanics is required to make recommendations for rehabilitation programs and prosthetic developments for lower-limb amputees. The thesis aim was to determine the biomechanical adaptations made by active unilateral trans-tibial amputees (TT) using a microprocessor-controlled ankle-foot prosthesis in active (MC-AF) compared to non-active mode (nonMC-AF) or elastically articulated ankle-foot device. A secondary aim was to determine the biomechanical adaptation made by able-bodied individuals when ankle motion was restricted using a custom made ankle-foot-orthosis and provide further insight into the importance of ankle dynamics when walking on ramps. Kinetic and kinematic data were recorded from nine TT's and twenty able-bodied individuals. Able-bodied participants, ankle restriction, led to an increase in involved limb loading response knee flexion that is accompanied by the increased knee power generation during the single-limb-support phase that correlates to TTs results. TT's use of an MC-AF reduced the 'plantar-flexion' resistance following foot contact allowing foot-flat to be attained more quickly. Followed by the increased 'dorsi-flexion' resistance which reduced the shank/pylon rotation velocity over the support foot, leading to an increase in negative work done by the prosthesis. These findings highlight the importance of having controlled ankle motion in ramp descent. Use of an MC-AF can provide TTs controlled motion for descending ramps and hence provide biomechanical benefits over using more conventional types of ankle-foot devices.

Keywords: Biomechanics, Gait, Amputee, Ramp descent, Ankle bracing, Microprocessor-controlled, ankle-foot prosthesis.

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Publications and Presentations

The list of publications that were completed during my doctoral studies at the University of Bradford and had been published in peer-reviewed journals is presented below. The data contained within this thesis was used for publications and presentations.

Publications

Struchkov, V. and Buckley, J. G. (2016) Biomechanics of ramp descent in unilateral trans-tibial amputees: Comparison of a microprocessor controlled foot with conventional ankle-foot mechanisms. Clin Biomech (Bristol, Avon) 32, 164-70. This work is assigned in Chapter 6 and 7.

De Asha, A. R., Barnett, C. T., **Struchkov, V**. and Buckley, J. G. (2016) Which Prosthetic Foot to Prescribe?: Biomechanical Differences Found during a Single-Session Comparison of Different Foot Types Hold True 1 Year Later. JPO: Journal of Prosthetics and Orthotics.

Presentation

Oral Presentation – ‘Effects of ankle-foot orthosis on joint kinetics during walking down slopes in healthy individuals.’ 14th Annual Staffordshire conference on clinical biomechanics, Stoke on Trent, UK, April 2016.

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Glossary

- ✓ TT – unilateral trans-tibial amputee;
- ✓ AB – Able Bodied;
- ✓ GRF – Ground Reaction Forces;
- ✓ vGRF – vertical Ground Reaction Forces;
- ✓ CoM –Centre-of-Mass;
- ✓ CoP – Centre-of-Pressure;
- ✓ A-P – Anterior-Posterior;
- ✓ M-L – Medio-Lateral;
- ✓ AFO- custom made Ankle-Foot-Orthosis;
- ✓ RoM - Range of Motion;
- ✓ BW – body weight;
- ✓ 6DoF - six (6) Degrees of Freedom;
- ✓ FJC - Functional Joint Centre;
- ✓ FP - Force Platform;
- ✓ SLS - Single-Limb-Support;
- ✓ IC - Initial Contact;
- ✓ TO - Toe Off;
- ✓ DS1 - Initial Double support;
- ✓ DS2 – Terminal Double support;
- ✓ SD - Standard Deviation;
- ✓ ANOVA - Analysis of Variance;
- ✓ MC-AF - a quasi-passive (active mode *Elan*) microprocessor-controlled hydraulically damped, uniaxial articulating ankle-foot device (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). Chapter 3.12;
- ✓ nonMC-AF - a hydraulically damped, uniaxial articulating ankle-foot device (non-active mode *Elan*) (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). Chapter 3.12;
- ✓ elastic-AF - elastic (rubber-snubber) (*Epirus*) (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). Chapter 3.12;
- ✓ VL – Virtual limb (defined as an angle between support, ankle functional joint centre and linked to the whole body Centre-of-Mass);
- ✓ *UDS* – unified deformable segment;

- ✓ DRF - dynamic response prosthetic-feet;
- ✓ ADL - activities of daily living;
- ✓ EWA – Early Walking Aid;
- ✓ IPOP- immediate post-operative prosthetic;
- ✓ SIGAM - Special Interest Group in Amputee Medicine;
- ✓ SSWS - self-selected walking speed;
- ✓ BACPAR- British Association of Chartered Physiotherapists in Amputee Rehabilitation;
- ✓ AMP - Amputee Mobility Predictor.

CHAPTER ONE - INTRODUCTION

1.1 Background

Loss of a lower-limb is one of the most psychologically and physically shocking incidents that can happen to an individual. The impact of lower-limb loss is profound for patients and their families. The amputation is also accompanied by a financial cost to NHS resources where the largest expenditure is due to prolonged stay, rehabilitation and long-term care (Singh et al. 1996; Moxey et al. 2010). The rehabilitation process is intended to return the lower-limb amputees independence; therefore facilitating the return to previously experienced activities of daily living (ADL). ADL involves the amputee’s ability to perform slope ambulation, change walking speed, approach stairs, etc. Throughout those tasks lower-limbs require adaptation to deliver safe and energy efficient movement. Where the prosthetic device functionality has an effect on biomechanical adaptations in overground gait (Underwood et al. 2004; De Asha et al. 2014) and ramp ambulation (Agrawal et al. 2015). Optimal performance of a task would require a prosthetic device that was able to change functionality accordingly to daily tasks.

A common ADL involves ramp ambulation (McIntosh et al. 2006). Ramp descent compared to overground gait involves control of body weight forward/downward transition (Smith et al. 1998; Lay et al. 2006). This control leads to an increased range of motion at the ankle and knee (Wall et al. 1981; Lay et al. 2006). To control the momentum generated by the fall of the body weight ankle dorsi-flexors that control plantar-flexion until foot-flat were utilised. The knee loading response absorbed the shock from the body weight fall from the contralateral limb. After attainment of the foot-flat, the ankle dorsi-flexors controls the forward rotation of the tibia over the foot to control gravitational energy when descending ramps (Saunders et al. 1953; Lay et al. 2006; Lay et al. 2007). The following push-off requirements are reduced due to the gravitational assistance in body weight transition (Lay et al. 2007). Therefore, lower-limbs have to adapt to ankle function during ramp descent.

For lower-limb amputees, the locomotor function is changed according to the constraints of the prosthetic devices used. Currently, the majority of prosthetic devices prescribed are not adaptable (Marinakis 2004). Where active unilateral trans-tibial amputees (TT) are frequently prescribed dynamic response prosthetic-feet (DRF), which could also be referred to as energy-storing and return prosthetic-feet. Basic DRF devices had a rigid 'ankle' and functioned only by deformation and the recoil of carbon fibre keels which simulates 'plantar-flexion' and 'dorsi-flexion'. A prosthetic ankle-foot which included elastic articulation at the point of attachment with a pylon has shown biomechanical benefits during ramp descent (Su et al. 2010). Nonetheless, those devices are non-adaptive and have set-up resistance of 'plantar-flexion and 'dorsi-flexion' according to overground gait with the self-selected customary speed of the amputee (Vickers et al. 2008). The use of those devices for adaptive gait could lead to discomfort (Klute et al. 2001) with compensations in weight-bearing joints and as a result, might lead to secondary physical conditions (Radin et al. 1973). Some of those compensations might also compromise gait safety. The National Health Service (NHS) England introduced a policy in July 2015 (Reference: NHS England D01/P/b) to approve the funding of Microprocessor Controlled Knees for those living with above knee, through knee and hip disarticulation amputations. To perform adaptive gait in a safe and efficient manner, prosthetic devices should ideally be able to change 'plantar-flexion' and 'dorsi-flexion' resistance according to the gait phase.

The thesis explored the recently developed and now commercially available advanced ankle-foot prosthetic device *Elan* (Chas. A, Blatchford and Sons, Basingstoke, UK). The MC-AF device is a quasi-passive microprocessor-controlled hydraulically damped, uniaxial articulating ankle-foot device that has to be 'tuned' appropriately to optimise each user walking speed and terrains. The prosthesis has incorporated a carrier that provides hydraulically damped articulation between the pylon and the DRF, which is designed for independent carbon fibre heel and fore-foot keels. This articulation within the deformation of keels under body weight load simulates 'plantar-flexion' and 'dorsi-flexion'. The *Elan* (Chas. A Blatchford and Sons, Basingstoke, UK) device during overground gait with the self-selected walking speed operates as a non-adaptable device *Echelon* (Chas. A Blatchford and Sons, Basingstoke, UK) but during ramp descent designed to adapt 'plantar-flexion' and 'dorsi-flexion' hydraulic 'ankle' articulation resistance to deliver safe and energy efficient locomotion. The *Elan* device, during ramp descent, after initial contact, reduces the hydraulic resistance of articulation to attain foot-flat sooner on the ground within the heel-keel deformation which simulates 'plantar-flexion'. When foot-flat was attained, the pylon rotates forwards where body weight transfers onto the fore-foot keel, to control the device increases hydraulic resistance with the keel deformation so simulates 'dorsi-flexion'. Therefore, the primary hypothesis of this research was that the articulation provided by the microprocessor-controlled hydraulic attachment (*Elan*) would allow reduced biomechanical compensations of the remaining lower-limb joints and improve safety of TTs during ramp descent compared to non-adaptable prosthetic ankles-feet: *Elan* in non-active mode nonMC-AF which behaves as (*Echelon*; Chas. A, Blatchford and Sons, Basingstoke, UK) and elastic-AF (*Epirus*; Chas. A Blatchford and Sons, Basingstoke, UK). The analogous prosthetic ankle-foot device 'Proprio-Foot' (Ossur hf, Iceland) is commercially available, although this device has doubtful biomechanical benefits during ramp descent (Fradet et al. 2010). Hence, a detailed investigation is critical to have a sound understanding of the underlying prosthetic device function.

This thesis should offer a deeper understanding about biomechanical adaptations when TTs using different functionality ankle-foot prosthetic devices during ramp descent. The primary aim of this thesis to investigate biomechanical changes when utilised *Elan* in active (MC-AF) mode compared to *Elan* in non-active mode (nonMC-AF) or elastic-AF articulated devices during ramp descent which would contribute to the development of device design improvement. Whenever ankle-foot device biomechanical advantages could be identified between prosthetic devices, then further recommendations could be made to prosthetic manufacturers. Another important aim was to deliver clinical recommendations to health practitioners for amputee rehabilitation within the improvement of gait safety. This would aid the return to their daily living activities. Nevertheless, to make recommendations for prosthetic manufacturers and health practitioners, it is important to have a deep understanding the underlying biomechanical function, where the investigation of the able bodied individuals with restricted ankle during ramp descent would develop this understanding. The use of the ankle restriction in able bodied individuals could simulate the effect of rigid ankle-foot device that is used by TTs. Indeed, TTs are distinct to able-bodied individuals when utilising an ankle brace due to limited proprioception and absence of distal muscular control. Although are required to determine whether biomechanical compensation existed in able bodied individuals between restricted and non-restricted ankle during ramp descent. Based on the current scientific papers, it is not clear how the ankle restriction affected able bodied individuals and what is crucial is the ankle function during ramp descent. Hence, the restriction of the ankle could provide further insight into the importance of ankle dynamics when descending ramps. The examination of the ankle with a custom made ankle-foot-orthosis (AFO) in restricted and non-restricted modes could also offer recommendations to health practitioners to improve the rehabilitation process for patients with lower-limb impairments. The use of a restricted ankle during ramp descent would lead to biomechanical compensations in the remaining lower-limb joints and/or would affect the gait pattern, which was the secondary hypothesis of this thesis. The combined examination of biomechanical adaptations by the remaining joints and/or alteration of the gait pattern in active TTs and able bodied individuals would add to the accumulative knowledge of the ankle function contribution in the biomechanics of ramp descent.

1.2 Structure of the thesis

Chapter one of the thesis includes the introduction, aims, and the central hypotheses. The primal aim of this thesis was to distinguish biomechanical alterations during ramp descent between prosthetic ankle-foot devices: *Elan* in active (MC-AF) mode, non-active (nonMC-AF) mode and *Epirus* (elastic-AF). The thesis has been divided into two parts. The first part deals with able-bodied individuals that utilised AFO unilaterally to perform ramp descent and overground gait. The used AFO has two ankle modes: restricted and non-restricted. The second part deals with active TTs, where the *Elan* was assessed in comparison to conventional; non-adaptable hydraulically and elastically articulated ‘ankle’ mechanisms. In the study, ankle-foot devices share the same carbon fibre fore-foot and heel keels; this was allowed to focus on the articulation between the pylon and tripod section (heel and split fore-foot keels). To aid the reader throughout the thesis the first chapter includes the terms of reference used.

Chapter two provides a detailed review of the significant scientific literature related to overground gait and ramp descent with specific attention on TTs. The review considers the drawbacks of each study with a possible solution for these methodologies. The initial part of the literature review begins with a brief overview of the lower-limb amputation epidemiology and trans-tibial amputations. The review is followed by an overview of commercially available lower-limb prosthetic developments on the market. The development of prosthetic foot designs is described from a rigid prosthetic ankle-foot to the quasi-passive microprocessor controlled ankle-foot device. The chapter includes the specifications of the prosthetic devices that were used in this thesis. The following areas focus on inverted pendulum theory, and how this relates to how the CoM (Centre-of-Mass) translates over the planted foot during locomotion. The literature review critically analysed the scientific papers which addressed able-bodied individuals and lower-limb amputees in overground and ramp descent gait. The literature review ends with the summary of analysed scientific papers. The chapter finishes with specific aims and objectives of this study. To investigate these devices was used a sequence of experiments, where a number of specific aims built to achieve objectives. Subsequent chapters provide a series of studies to investigate the current gaps in the literature as defined by the thesis objectives. The specific aims and objectives were investigated in this chapter.

Chapter three includes ethics and inclusion criteria of able-bodied individuals and active TTs with a description of the classification of activity level. The chapter includes a general methodology that was used for experimental chapters and describes the equipment used in this thesis. The following experimental chapters include specific methodology details of experiments, results and discussion/conclusion. To test the difference between conditions in all experimental chapters a repeated measures design was used.

Chapter four (first experimental chapter) and chapter five (second experimental chapter) investigated the effects of the use by able-bodied individuals of a custom made ankle-foot orthosis with two ankle conditions: restricted ankle and non-restricted ankle during overground and ramp descent. Chapter four focuses on the sagittal plane, whole body dynamics within joint kinematics and the spatio–temporal parameters of the involved (with AFO) and non-involved limbs. It was hypothesised that during single-limb-support Centre-of-Mass relative to the ankle of the support foot would increase the angular velocity with restricted ankle otherwise it would increase the knee flexion.

Chapter five investigated the support (involved and non-involved) limb joints kinetic compensations to restricted ankle for ramp descent in comparison to overground gait. The main hypothesis of this chapter is that restricted ankle affected ankle ‘push-off’ positive work done during stance, but in ramp descent, with gravity assistance, this would reduce the anticipated increased negative work during initial double-support and single-limb-support. Hence, as compensation, there would be increased knee (with AFO) involvement during these periods.

Chapter six (third experimental chapter) and the chapter seven (fourth experimental chapter), examined TT with two different ankle-foot prosthetic devices: *Epirus* (elastic-AF) and *Elan* in active (MC-AF) and non-active (nonMC-AF) modes during ramp descent with self-selected walking speed and comfortable slow walking speed. This chapter has been divided into three parts; the first part deals with ways to investigate the effects of the microprocessor-controlled hydraulically damped ankle-foot device and whole body dynamics, lower-limb joint kinematics' the second part deals with the Centre-of-Pressure forward velocity and the third part investigated the effect of prosthetic ankle-foot articulation types on spatio-temporal symmetry of the gait. It was hypothesised that *Elan (*MC-AF) device in active mode compared to *Elan* in non-active mode (nonMC-AF) or *Epirus* (elastic-AF) ankle-foot device would reduce whole body momentum (angular velocity) during single-limb-support and/or reduce residual-knee flexion with shank angular velocity. On the other hand, Centre-of-Pressure forward velocity during single-limb-support would be reduced with such a device compared to non-adaptive (nonMC-AF and elastic-AF) articulated prosthetic devices and so improve dynamic stability during ramp descent.

<u>Chapter seven</u> begins by laying out the theoretical dimensions of the research and looks at how ankle-foot prosthetic devices would impact ramp descent with two speeds. Chapter seven determined whatever use of *Elan* in active mode (MC-AF) compared to *Elan* in non-active mode (nonMC-AF) or *Epirus* (elastic-AF) would improve the kinetic of ramp descent. The chapter provides an examination of differences between prosthetic device moments and powers as a unified deformable model. In addition, the chapter examines the effects of ankle-foot articulations on GRF of the contralateral side. It was hypothesised that a microprocessor-controlled hydraulically articulated 'ankle' device would attain foot-flat quicker than that followed by a reduction of shank/pylon forward rotation compared to conventional articulated 'ankle' mechanisms and as a result, the use of such a device would reduce knee flexion and mechanical power in early stance.

<u>Chapter eight</u> provides a discussion, finalising all experimental chapters and the relationship between the current thesis and other researches have been investigated. The chapter also contains thesis limitations and recommendations for future studies based on the thesis findings and current literature. Finally, the chapter closes with a summary of the thesis that specifies knowledge on the

biomechanical differences during ramp descent between ankle conditions (restricted/non-restricted) in able-bodied individuals and between ankle-foot articulations (*Elan* active mode (MC-AF), *Elan* in non-active mode (nonMC-AF) or *Epirus* (elastic-AF)) in TTs. The findings could have a positive effect on rehabilitation, physiotherapy treatment and lower-limb prosthetic development.

CHAPTER TWO - LITERATURE REVIEW

2.1 Epidemiology

The word amputation was mentioned for the first time in Roman texts but referred to cutting off the hand as a punishment and did not mean surgical removal. The first recorded surgical removal of a limb necessary for life preservation was performed and reported by Hippocrates between 460-370 BCE (Kirkup 2007). The basis of this technique persists in current surgical practice with the added improvement of anaesthesia, haemostasis and preoperative procedures. Major improvements in amputation aetiology took place in the last century during World War I and II. At present, the causes of amputation are often the consequence of unsuccessful treatment and are performed due to various causes: vascular disease, malignancy (tumours), congenital deficiency, or severe trauma (Dillingham et al. 2002).

Loss of a lower-limb is one of the most physically shocking incidents to happen to individuals and is also associated with a high financial cost. The full cost of lower-limb amputees surgery, rehabilitation, hospitalisation physical therapy and prosthesis can exceed $50,000 (US dollars) in the United States of America (Green et al. 2001). In Australia, an amputation costs around A$12,815 (Australian dollars) (Davis et al. 2006) and in the UK for the National Health Service (NHS) between £10 -15 000 (UK pounds) (Moxey et al. 2010). This amount continues to rise because studies estimated these figures during the last decade. To return patients to their previous level of function requires the correct rehabilitation process which involves significant financial cost. The rehabilitation process is continuing to peruse the development of training and equipment. Use of the appropriate training and equipment facilitates the performance of different daily tasks in a safe and energy efficient manner and aids development of patients independence with the potential for a reduction in the cost of care.

A lower-limb amputation also has a psychological effect on the patient. Psychological effects could include factors such as depression and anxiety, social functioning and discomfort, body image anxiety, sense of self-identity about physical limitations (Horgan and MacLachlan 2004). The psychological symptoms like depression may also be associated with the degree of functionality available with a prosthetic device (Singh et al. 2009). Therefore, the selection of the correct rehabilitation process with the appropriate prosthetic device should help the patient to return to their previous lifestyle.

Approximately 20% of adults older than 55 years old have been affected by peripheral vascular disease (PVD) (the result of narrowing or blockage of the arteries) in Europe and North America, which leads to 70-75% of lower-limb amputations (O'Donnell et al. 2011). The United States (US) statistics indicated that 664,000 people had a major lower-limb amputation in 2005 and an estimated 3.6 million people will be living without limbs by 2050 (Ziegler-Graham et al. 2008; Varma et al. 2014). There are 40 thousand trans-tibial amputations performed annually in the US (Dillingham et al. 2002). The United Kingdom (UK) NASDAB (National Amputee Statistical Database, UK) data have presented around 5,000 lower extremity amputations annually during the 2005-2011 period. The most recent annual Limbless Statistics (2011/12) stated 5387 lower extremity amputations within around 10% of traumatic cases. More than half (56%) of total amputations were trans-tibial (the loss of the ankle joint below the knee through the tibia). In total, traumatic cases were around 11% of trans-tibial amputations. The majority of traumatic amputations are due to road traffic accidents (NASDAB, UK, 2011/12). In addition to this are individuals that lost their limbs during military service. The British military identified that 21 individuals became TTs during the period March 2004 - March 2010 (Bennett et al. 2013) and there were 683 trans-tibial amputations in the US service in a period of ten years (January 2001 - July 30, 2011) (Krueger et al. 2012). Formerly active individuals are likely to lose a limb due to trauma: road traffic accident; sports injury or military activities. The number of traumatic amputations increases with the rise in the usage of high-powered vehicles and improvements in resuscitation techniques (Henderson et al. 1982).

International health practitioners recommend lower-limb amputee patients accommodate a healthy and active lifestyle (*a lifestyle that contributes positives to physical, mental and social well-being and includes regular exercise*) that would improve their health-related quality of life (Medhat et al. 1990; Waxman and World Health 2004). Although, in the research of Deans and colleagues has presented weaker than the expected relationship between physical activity and quality of life (Deans et al. 2008). Nevertheless, the majority of patients are determined to maintain the activities of independent daily living (ADLs) and to be an active member of society. The ADLs are activities that are required for normal self-care. Those activities are defined as: personal hygiene, movement in bed, transfers from one seat to another and changing position from sitting to standing, dressing, eating, bowel and bladder control and locomotion. In this case, locomotion is not only the ability to perform overground gait but also to accomplish different tasks such as walking on different gradient slopes or using stairs. These tasks are an important part of an active individual in today's lifestyle. Unfortunately, the residual limb has reduced proprioception, and prosthetic devices do not have the same functionality as a healthy limb. Over 68% of amputee patients wear a prosthesis at least seven hours a day to fulfil their everyday activities (Pohjolainen et al. 1990). The long-lasting use of the prosthesis could lead to secondary physical conditions that include osteoarthritis, osteoporosis, back pain, and other musculoskeletal problems. This is likely to be the result of increased forces on the contralateral side within alteration of the biomechanics of gait so lead to complications that negatively affect the gait and of amputee patients (Nolan and Lees 2000; Nolan et al. 2003). Hence, a prosthetic device design should mimic the human limb function during different tasks to restore the previously experienced lifestyle. Seeing that, the prosthetic device functionality was associated with quality of life (Gallagher and Maclachlan 2004). To fulfil lifestyle requirements to perform tasks such as walking on different gradient slopes or using stairs requires adaptation of the ankle function to provide safe, comfortable and energy efficient locomotion. The use of prosthetic devices would involve compensations by the remaining joints, according to the tasks performed and the device functionality. The use of conventional (non-adaptable) prosthetic devices during these tasks could affect safety and/or excessive compensations by remaining joints as it set up for overground gait with the self-selected walking speed (Vickers et al. 2008). Nevertheless, amputee patients had to have previously experienced these tasks. Thus, it can be suggested that enhanced functionality of the prosthetic device has to be designed, prescribed, and used by active amputees. Furthermore, the literature review will relate to studies with active amputees.

2.2 Lower-limb amputations

Lower-limb amputation levels can be categorised as the minor or distal and major or proximal. The UK Limbless statistical data base has presented that in 2011/12 were minor lower-limb amputations performed around 2% of compared to around 98% of major (NASDAB, UK, 2011/12). The minor amputation performed by a ‘cut-off’ toe or part of a foot and defined below the ankle joint (a specific code to identify a procedure: ICD9-CM: 84.11–84.12) with restriction to the toe or partial foot (Lombardo et al. 2014). Amputation of toes could be partial, complete or disarticulation (at the metatarsophalangeal joint) and ray (toe and metatarsal). The big (great) toe is considered to have the most contribution into locomotion compared to other toes (Hughes et al. 1990), so the functionality of the patient would be dependant on the level of amputation. However, the level of the toe amputation is dependent on the degree of disease and applied the surgical technique. The proximal foot amputation includes transmetatarsal, tarsometatarsal (Lisfranc disarticulation), midtarsal (Chopart disarticulation) (Ploeg et al. 2005; Apelqvist et al. 2008). Partial or complete toe amputation (minor or distal) would lead to insignificant reduced mobility and generally can be fixed by shoe corrections (Wagner 1981). Minor amputation does not necessarily require a prosthesis for locomotion. On the other hand, the major or proximal lower-limb amputation would require a prosthetic device to perform relatively natural locomotion. This amputation is typically defined as above the ankle joint (a specific code to identify a procedure: ICD9-CM: 84.13–84.19) (Lombardo et al. 2014). Major amputations is divided into categories: hemipelvectomy (through pelvic bone), hip disarticulation, trans-femoral (above the knee), knee disarticulation (at knee joint or Gritti-Stokes amputation), trans-tibial (below knee), ankle disarticulation (at the ankle joint). The majority of the UK limbless population in 2011/12 accounted for by 56% of trans-tibial and 38% of trans-femoral referrals from all lower-limb amputations (NASDAB, UK, 2011/12). Trans-tibial over trans-femoral amputation is a noticeably reduced perioperative mortality (Bates et al. 2006). The healing rates after trans-tibial amputation is over 75% compared to partial-foot amputations with only 50% (Dillon and Fatone 2013). A more proximal level of lower-limb amputation increases the energy consumption (Waters et al. 1976) with decreased walking velocity (Genin et al. 2008; Vllasolli et al. 2014). These amputations are performed in order to retain a distal joint with the appropriate residual limb length for a prosthetic socket. This would provide the best chance of staying mobile after the surgery. Nevertheless, it has been shown, retaining ankle joint with partial-foot amputations compared to trans-tibial did not show a better balance (Kanade et al. 2008), energy cost or has reduced a compensatory strategy during locomotion (Dillon and Fatone 2013). Prosthetic foot devices could provide functionality comparable to an ankle with the partial foot. Hence, a more proximal level of lower-limb amputation is more critical for patients with major or proximal amputations.

The most common major lower-limb amputations in the UK are trans-tibial (NASDAB, UK, 2011/12). Trans-tibial or below knee amputation is known as the surgical removal above the ankle joint and below the knee joint (through the tibia) (Lexier et al. 1987). Trans-tibial amputation is performed when treatment of a foot or ankle has failed. This failure could be the result of severe injury, a severe infection, poor blood flow to the limb, non-healing ulcers, loss of function to the limb, birth imperfections, tumour and others sources of chronic limb pain. In some causes of amputation could be affected by time; then amputation should be performed urgently. For example, an increase of ischaemic time could lead to serious muscle loss by necrosis (Khalil and Livingston 1986), so prompt discussion should be conducted between the patient and orthopaedic surgeon. The optimal preservation of the residual limb length is aided by an appropriate fitting of the prosthetic socket. The surgical trans-tibial amputation procedures have guidelines according to the British Association of Plastic and Reconstructive Surgeons (BAPRAS) (www.bapras.org.uk).

The ideal residual tibial length for a patient with 180 cm height was offered by Pinzur (Pinzur et al. 2007) between ~10 and 18 cm. Although, the BAPRAS guideline was narrowed to allow between ~15-17 cm, below the knee joint of a tibia bone measured from the medial tibial plateau to the distal end of the tibia. On the other hand, the historical origins of the trans-tibial amputation, surgeons used a palm width ~10-15 cms. The fibula bone is resected 1-1.5 cm proximal to avoid contact with the end of the residual limb (www.bapras.org.uk). To aid coverage of the remaining bones, the muscle of the gastrocnemius and soleus must be resected to provide a more manageable flap. The flap is secured to the anterior tibia by a suture which provides the comfortable fitting of a prosthetic socket. If the residual limb length below the knee is too long, it could lead to a lack of muscle tissue that provides cushioning with the prosthetic socket as well as a blood circulation problems, so could increase the skin breakdown which would lessen long-term success (Levy 1995). Another critical disadvantage of a long residual limb is limited space available to fit the socket with a prosthetic foot. In contrast, if the length of the residual limb is too short, it possibly causes complications with the fitting of a socket. As a result, the reduced length (lever arm) and contact profile between the residual limb and the socket. This may also have a slight influence on the energy expenditure of the gait (Gailey et al. 1994). The shortest rational length has to preserve the functionality of the knee (flexion/extension). Hence, the optimal residual limb length has to provide a balance between weight bearing (comfortability) and torque (link with the prosthetic socket). The length could be defined by the elimination of the shortest and longest residual limb length. A patients height and residual limb length after trans-tibial amputation have determined the prosthetic foot device that could be used (Powelson and Yang 2012). The advanced prosthetic feet have increased build prosthetic height (Laferrier and Gailey 2010) and would not be suitable for patients with the long residual limb.

Trans-tibial level of amputation compared to trans-femoral has reduced energy consumption required for gait (Waters et al. 1976) and increased walking speed with decreased oxygen/energy consumption (Huang et al. 1979; Genin et al. 2008; Vllasolli et al. 2014) as the result of the preserved knee joint. Overall, the effects indicated that the significance of the correct level of amputation could not be overemphasised due to the significance of the impact on amputees' gait. Nevertheless, the length of the residual limb should preserve the distal joint, but only if aetiological factors and clinical examination allow it. The length of the residual limb has to be preserved to the length with disease eradication but have to consider an optimal connection with socket (interface) to provide the functionality of the prosthetic device (Grevsten and Erikson 1975). Critically, the amputation surgery has to provide comfortable, painless weight bearing through the residuum. Nevertheless, amputees have reduced weight bearing through the end of the residual limb (Breakey 1976; Engsberg et al. 1991). The orthopaedic surgeons have to take into consideration the healing process of the wound beside the post-surgical function of the prosthetic side to provide prompt rehabilitation. Prompt weight bearing mobility for the amputee helps to avoid deconditioning and permits early discharge from hospital and further quicker improvements during rehabilitation program.

2.3 Postoperative mobility

Following lower-limb amputation patients aim to restore and maintain a certain level of mobility. This improved mobility offers independence with a better quality of life for the patient as well as reducing health care cost (Davies and Datta 2003; Asano et al. 2008). To regain mobility after amputation, patients commonly follow a specific rehabilitation program. During this program, the patient is required to re-learn locomotion with different prosthetic components in order to return to their previous lifestyle. Considerations of this program include patients' level of pain and costs involved. In the UK, the recommendations in postoperative rehabilitation for TT amputees are provided by the British Association of Chartered Physiotherapists in Amputee Rehabilitation (BACPAR). The prompt rehabilitation critically depends on the residuum postoperative management in order to achieve residual limb functionality (Nawijn et al. 2005). There, the functionality of residual limb influences postoperative gait biomechanics.

The key to creating a functional residual limb and providing proficient prosthetic control is to carry out the correct postoperative healing of the wound and oedema reduction. Oedema occurs after surgery as a natural result of damage to the tissue during the surgery, accident or interference with the tissue fluid transfer mechanism (Redhead and Snowdon 1978; Janchai et al. 2008). Persistence of oedema could cause a delay in the final rehabilitation. To prevent excessive oedema (Golbranson et al. 1968; Gerhardt et al. 1970) requires appropriate wound management techniques. The technique has to control oedema and be safe and easily applied in order for the wound to remain secure. Compression therapy is commonly used to reduce oedema (Condie et al. 1996), which utilises: shrinker socks, crepe bandages, Elset™ bandages, plaster casts, etc. (Condie et al. 1996). The dressings used are termed as a soft, rigid dressing, semi-rigid dressing, silicon, and gel-liners.

Historically, a soft dressing was commonly used during World War I (1914-1918) (Smith et al. 2004) and is still utilised by some orthopaedic surgeons. The dressing consists of sterile gauze and fluff which is commonly followed by a compressive elastic wrapping bandage to fasten the soft gauze and control oedema (Baker et al. 1977; Choudhury et al. 2001). Advantages of soft dressing use include wide availability of materials used and its low cost, easy application and short time required to apply it, which would allow a wound to be examined more frequently. On the other hand, the use of soft dressing on a freshly amputated residual limb could mean strong pain for the patient during locomotion, decelerate the healing process and delay the start of weight bearing, this could, in turn, lead to a prolonged stay in hospital with the increase in health care cost (Weinstein et al. 1988; Smith and Fergason 1999). Besides that, a soft dressing followed by application of elastic wrapping bandage can produce excessive pressure, so can lead to tissue necrosis as well as the possibility of infections, bruising, wound breakdown burns, or ulceration (Troup 1988; Smith et al. 2004) as well as knee flexion contracture. Later on, to improve patients postoperative wound management a rigid dressing was introduced by Berlemont in 1958 (Berlemont 1961). The application of thigh level rigid cast dressings commonly begins with a soft gauze dressing then a plaster cast is rolled and moulded (Jones and Burniston 1970). Rigid dressing allows an early walking aid to be employed (EWA) and to start gait training sooner (Golbranson et al. 1968; Baker et al. 1977; Wu et al. 1981; Nawijn et al. 2005). Patients with rigid dressings have a few weeks delay in gait training if used without an immediate prosthesis (Golbranson et al. 1968). Further design updates introduced a rigid dressing which incorporated an immediate postoperative prosthetic (IPOP) (Moore et al. 1972; Weinstein et al. 1988). The design had a rigid dressing cast that immediately attached to the postoperative prosthetic foot which allowed residuum weight bearing in 12 hours (Smith et al. 2004). An alternative modification of the rigid dressing with IPOP that allows Wu and colleagues presented knee flexion as a short removable rigid dressing (Wu et al. 1979; Wu and Krick 1987). This dressing was shorter than the thigh level dressing and combined a rigid dressing polyvinyl-chloride pipe to model a pylon component attached to a preparatory prosthetic foot with a prosthetic sock under the cast (Wu and Krick 1987). The technique allowed the use of pre-fabricated prostheses and assisted patients to move in less than 2 hours. A similar IPOP technique utilises thigh level prefabricated pneumatic prostheses or below the knee (Little 1970; Pinzur et al. 1989; Schon et al. 2002). To surround the residuum in a prefabricated pneumatic prosthesis utilised air cells, which cover an area of the socket or an airbag system which includes a single plastic prosthetic component that fits over one or more pneumatic airbags (Rheinstein 2000; Schon et al. 2002; Reichmann et al. 2017).

Rigid compared to soft dressing utilisation included advantages such as: prompt oedema reduction with healing acceleration, reduction of the requirement for revision surgery (Mooney et al. 1971; Pollack and Kerstein 1985; de Noordhout et al. 2004; Smith et al. 2004; Nawijn et al. 2005), progressive residuum shrinkage, reduction of skin breakdown, pain reduction (Reichmann et al. 2017) and, improves weight bearing acceptance. Early weight bearing allows the rehabilitation process to begin sooner, reduce rehabilitation time (Kraeger 1970; Nawijn et al. 2005) and facilitates the prompt utilisation of a functional prosthesis while gradually re-learning locomotion (Dickstein et al. 1982; Folsom et al. 1992; Scott et al. 2000; Broomhead et al. 2003; Vanross et al. 2009; Ali et al. 2013). The immediate start of the rehabilitation process after amputation allows patients to obtain mobility, independence and safety, which are the quality of life factors (Millstein et al. 1985; Sheikh 1985). This, could in turn, reduce mortality, lessen the risk of more proximal amputation, contribute to greater medical stability, and prosthesis acquisition (Dickstein et al. 1982; Condie et al. 1996; Dillingham and Pezzin 2008). However, postoperative mortality is more likely associated with above the knee rather than below knee amputees (Lim et al. 2006; Basu et al. 2008). Also, the use of the EWA or IPOP devices has shown a positive influence on patient psychology as a result of the change in focus from the limb loss to recovering and achievement of the previous level of activity (Smith and Fergason 1999). Patients commonly feel that re-learning to walk as soon as possible is a crucial transition for returning to their previously experienced lifestyles within physical and social activities. The use of the EWA or the IPOP devices presented advantages in postoperative wound management. However the patient's physical/medical state, previously experienced lifestyle and cognition have to be taken into consideration.

The regularly prescribed EWAs for trans-tibial amputees is the Pneumatic Post Amputation Mobility aid (PPAM) (Redhead et al. 1978) or the Amputee Mobility Aid (AMA) (Scott et al. 2000). PPAM devices are widely available and have relatively low-cost (Sher and Liebman 1982; Reith and Arneja 1992) due to simple design (Reith and Arneja 1992; Scott et al. 2000) with a fixed residual knee in a relatively extended position. The AMA design was developed later, that allowed the knee flexion and extension by using a hinge mechanism. Both EWA designs are equipped by pneumatic bags (Sher 1974; Sher and Liebman 1982; Rausch and Khalili 1985; Scott et al. 2000; Schon et al. 2002) to control excessive long-term pressure, which could affect tissue damage and delay the wound healing process. Nevertheless, both designs have shown similar pressure and pressure fluctuations during supported walking between residuum and the pneumatic bag (Scott et al. 2000). In the study Barnett et al., PPAM and AMA users have shown an increase in walking speed as well as a change of gait pattern throughout the rehabilitation process (Barnett et al. 2009). During early rehabilitation, both groups with functional prosthesis demonstrated an increase in step length when the lead is the intact limb compared to prosthetic lead (Barnett et al. 2009). This is crucial as it would improve gait symmetry because step length is shorter for TTs patients when the lead is the intact limb (Isakov et al. 1996b; Mattes et al. 2000; Barnett et al. 2009). The examination of lower-limb joint kinematics and basic parameters of amputees' gait pattern in the study of Barnett et al. 2009 does not show differences between EWA groups at discharge after the rehabilitation program (Barnett et al. 2009). The study has examined amputee gait throughout five different points in the rehabilitation process. However, the examination does not consider the forces that affect this joint motion. Hence, the examination of joint kinetics could present deeper insights into the rehabilitation process and differentiate effects between EWA designs in the patient locomotion. Indeed, the number of participants is critical as it could affect statistical outcomes, in the study a small and uneven number of participants was included (AMA n=8; PPMA n=7) (Barnett et al. 2009). The examination of a larger and even number of participants could add to the knowledge of rehabilitation.

For successful and prompt recovery from the amputation surgery, it is crucial to inform patients about the future rehabilitation program (Smith and Michael 2004) with a visit to a prosthetist. The purpose of the visit is to offer pre-prosthetic management in order to accelerate the maturation of the residual limb among a further rehabilitation process that involves re-learning to walk with EWA then transfer to the functional prosthesis. To promote gait development during early rehabilitation stages, patients use an upper extremity for support on parallel bars to assist with transferring and reduce weight bearing. An appropriate rehabilitation process continually promotes gait adaptation for optimising weight transfer. After establishment of gait pattern, patients progress from the parallel bars to crutches and then to unilateral support. After establishing ambulation on overground surfaces, patients approach stairs, curbs, ramps, and uneven terrains. To improve ambulation, the patient has to perform exercises on flexibility, muscle strength, cardiovascular training, and balance under the supervision of a physician (Esquenazi and DiGiacomo 2001). Throughout the different phases of the rehabilitation process patients continuously adapt the biomechanics of gait (Barnett et al. 2009). Even after discharge from rehabilitation amputees still continuously adapt gait strategies according to the prosthetic device used as well as ADL tasks approached (Pezzin et al. 2000).

2.4 Classification of mobility

The decision about the functional prosthetic device prescription made conjointly by the patient and the rehabilitation team of physician, physiotherapist, prosthetist, specialist nurses and may also include the services of an occupational therapist and psychologist depending on the patient requirements. The choice of functional prosthesis prescription is based on matching the mechanical characteristics of the prosthetic device with the functional capabilities of the lower-limb amputee (Cortés et al. 1997). To define the lower-limb amputee functional capabilities requires suitable, reliable and valid clinical measures of the patient’s mobility. There are currently a number of mobility grading systems in use, so a review of the mobility grading systems that are employed in the clinical setting is required.

The most commonly used system of grading to define amputees’ mobility is a self-report questionnaire, the Special Interest Group in Amputee Medicine (SIGAM) is employed by the Consultant in Rehabilitation Medicine and the Multidisciplinary Rehabilitation Team at the limb fitting centres. SIGAM was introduced by Ryall and colleagues (Ryall et al. 2003). The SIGAM was created alongside the Harold Wood Stanmore grades (Hanspal et al. 1991) and is sensitive to the amputees’ mobility change. The SIGAM has six grades from A to F, where A grade is the lowest (non-prosthetic limb users), and F grade is the highest (normal or near normal walking) activity level. The scale is designed to facilitate grade assignment of the amputee mobility assesses with their habitual prosthesis. The result of a self-report questionnaire leads to the selection of the amputee mobility grade in clinical settings (Ryall et al. 2003; Rommers et al. 2008). The SIGAM questionnaire uses distance and the utilisation of walking aids, to define the user grade. For validity and sensitivity to change the SIGAM mobility grades examined compare to valid mobility measures as the timed walking test (TWT) (Wade 1992) and the Rivermead Mobility Index (RMI) (Collen et al. 1991). Examination of the SIGAM scale has shown validity and reliability with an effect size of 10.66 and the inter-rater reproducibility overall Kappa value of 0.86. The SIGAM grade system is restricted to identifying mobility regarding the help required to mobilise, distance, the use of walking aids, ability to negotiate difficult terrain and weather conditions.

Another commonly used self-report questionnaire is the Locomotor Capabilities Index (LCI) this is a validated measure to identify the mobility level of lower-limb amputees with their habitual prosthesis (Gauthier-Gagnon et al. 1998). The LCI is used by the clinicians to measure amputee's ability to operate the prosthesis that designed for the particular level of activity (Franchignoni et al. 2004; Larsson et al. 2009). An example of the LCI questionnaire is presented in Appendix 1. The LCI includes basic and more advanced daily tasks to identify locomotor capabilities of the amputee with their habitual prosthesis. Each task is scored on a 4 level scale with a maximum score of 56. The tasks are designed to identify the activity level of a patient so an appropriate prosthesis could be prescribed. According to the LCI questionnaire, less active patients would be prescribed prosthetic devices that provide the required functionality during their activities. This suggests, that patient’s LCI score should present his/hers level of activity according to this score with the corresponding functionality of the prosthetic device.

The US Health Care Fining System adopted the Medicare Functional Classification Level (MFCL) (HCFA 2001). Interestingly, MFCL is not based on any formal scientific research. The MFLC has five levels (K0, K1, K2, K3, K4) of the functional classification system. The number is used to identify the current state of activity level in assessing lower extremity amputees to prescribe an appropriate prosthetic device. The identification of the K level of lower extremity amputees activity is specified by the subjective discernment of physicians and prosthetists, so commonly relies on clinical measurements at the time of the examination. The examination could be performed on both the capacity and potential of the amputee (Hafner and Smith 2009). The MFCL is a tool to define a functional state that expresses the medical needs of amputees for particular prosthetic components to deliver necessary functionality. To make decisions on the amputee's condition, physicians and/or prosthetists consider the previous activity before amputation, past and current health, residual limb condition, associated medical problems, and the amputee desire/motivation for activity and other factors. The patients with activity level 'K0' do not have mobility so a prosthesis would not improve mobility or lifestyle. This base level is assigned to amputees who do not have the ability or potential to ambulate or transfer safely with or without assistance. Amputees 'K1' have very limited mobility level of activity. Those amputees have the ability or potential to use a prosthesis for transfers or ambulation in overground at a fixed walking speed. Following the level of activity 'K2', the patients have limited mobility, so the amputee has the ability or potential to use a prosthesis for ambulation and the ability to adjust for low-level environmental barriers such as kerbs, stairs, or uneven surfaces. The lower extremity amputees with activity level 'K3' have basic to normal mobility. Those amputees have the ability or potential to use a prosthesis as basic ambulation with the ability to adapt gait to most environmental barriers and alter walking speeds. The highest or advanced level of activity for lower extremity amputees, 'K4' was applied to amputees with bilateral involvement, active adults who exceed the basic use and also athletes. A similar system used in England to assist prosthetic limb services: (A) activity code with classification from A0L to A4L and specific sports limbs (Extra Contractual Activity). Unlike the SIGAM or the LCI, the MFCL system has the capability to define amputees with an upper level of activity due to the ability to vary walking speed and the ability to overcome environmental barriers.

The recent pilot study has shown, that amputees of K2 and K3 mobility level are comparable with the Amputee Mobility Predictor (AMP) measures, but K1 and K4 require further verification (Dillon et al. 2017). The AMP is a performance-based functional assessment tool that provides an objective measure of the ambulatory potential during and after rehabilitation and predicts the function of the following prosthetic prescription in lower-limb amputees (Gailey et al. 2002; Gailey 2006). The design of the AMP verifies the amputee’s readiness for ambulation. The AMP requires appropriate training for rehabilitation professionals to introduce the use of the AMP with reasonable confidence. This method includes 21 tasks in 6 different fields: sitting balance, a transfer from chair to chair, standing balance, gait, stairs, use of assistive devices. All tasks performed with the patient's habitual prosthesis. To validate the research Gailey et al. (2002) used patients with mean age of 54 years old which is relatively young for the lower-limb amputee population, as more than 73% of the UK amputee population is aged over 54 years old (NASDAB, UK, 2011/12). This is the drawback of the method as it was validated by the participants with a mean age of 54 years old, so the performance of some the AMP tasks for less mobile patients can be problematic (Gailey et al. 2002). In addition to this older amputees are frequently dysvascular and often have a low level of activity, so may not be able to perform these tasks.

A number of amputee mobility measures have been employed in the prosthetic rehabilitation settings. Prosthetic associated health professionals use a self-report questionnaire that includes not only described above in the SIGAM and the LCI, but also many other measures such as: the prosthesis evaluation questionnaire (PEQ) (Legro et al. 1998), the prosthetic profile of the amputee (PPA) (Grise et al. 1993), the orthotics and prosthetics users' survey (OPUS) (Heinemann et al. 2003), the trinity amputation and prosthesis experience scales (TAPES) (Gallagher and Maclachlan 2004), and other research groups. Certain systems have shown validity and reliability, whereas others have a lack of sensitivity (Condie et al. 2006). There is no definitive agreement between clinicians about the standards for the valid selection of prosthetic devices that can support the needs and abilities of the patient (Callaghan and Condie 2003; Sagawa et al. 2011). Currently, the preference of lower-limb prosthetic components is based on subjective knowledge and experience of the rehabilitation team (Schaffalitzky et al. 2011). To assess the amputee's level of mobility, often, clinicians preferred a self-report questionnaire due to the ease of use, and it displays the patient's perspective. These questionnaires can assess only limited points of mobility (Rommers et al. 2001), and this type of assessment does not have sufficient sensitivity and cannot define amputees upper level of activity (Pasquina et al. 2006). Therefore, clinicians often employ self-report questionnaires such as the SIGAM and the LCI mid to lower activity amputees. To confirm a self-reported questionnaire, it could be used in conjunction with other tests, such as the timed walking test (TWT) (Wade 1992) or the Timed Up and Go tests (TUG) test (Schoppen et al. 1999). However, the use of a self-report questionnaire on amputees with the upper level of mobility cannot distinguish between their levels of mobility. Besides that, the answers of most amputees will be affected by their willingness to return to a previously experienced lifestyle, receive improved prosthetic fittings, and the level of prosthetic care. The examination of a patient's performance during a task or a number of tasks can be used independently. The performance-based measures are scored according to the time to complete the task, the distance covered, or the amputee's capability to do the task. Except for mention above the AMP, the TWT and TUG tests clinicians also can employ tests as the comprehensive high-activity mobility predictor (CHAMP) (Gailey et al. 2013), the six-minute walk test (6MWT) (Balke 1963) and others. Certainly, more advanced biomechanical measures such as kinematic, kinetic, and temporal-spatial parameters can be used to identify amputees' level of mobility. It is not always possible to measure these parameters in clinical settings, but the development of recent technologies can support clinicians to measure the required biomechanical parameters. To detect differences between prosthetic feet measure work symmetry between the lower-limbs as effort, delivering by each limb during gait utilised the symmetry in external work (SEW) (Agrawal et al. 2009) is used in sole sensors. Nevertheless, clinicians are highly recommended to standardised measures to examine the effect of a prosthetic on the patient as a current clinical professional unsure what the best prosthetic components is for the particular amputee populations (Deathe et al. 2002; Gailey 2006). Optimum correspondence when the measures of the patient mobility level based on the functionality of prosthetic components.

2.5 Prosthetic developments

2.5.1 Evolution in lower-limb prosthetics

Lower-limb prosthetic devices have evolved with time from basic to advanced and sophisticated versions. The first use of prosthetics was mentioned in the 18th dynasty of ancient Egypt although the first real rehabilitation prosthesis was mentioned in Greek and Roman civilisations, these were typically made from bronze or copper with a wooden base and leather straps for attachment to the residual limb (Thurston 2007). For centuries, a leather corset, or lacer would have been used to attach handmade wooden prosthetics (wooden peg) for TTs. The majority of prostheses at the time were relatively durable but very heavy. A form of corsets was distally opened and fastened the prosthesis to the residual limb, as the accumulation of body fluids was frequently present at the residual limb. The evolution of lower-limb prosthetic devices has reached sophisticated designs that help amputees to achieve a more efficient and safer gait. Currently, a trans-tibial prosthesis consists of a socket, suspension, pylon-shank and foot. The socket of the device has to be made to exactly fit the residual limb to provide an extension of the residual limb. Other components are modular and could be individually assembled for a particular patient’s parameters to achieve optimal functionality. The modular prosthetic is normally made of lightweight materials such as aluminium and titanium. The modular construction also allows parts to be independently exchanged in addition to servicing the device. The developments of polymer technologies, to produce durable and light fore-foot and heel keels were using carbon fiber. Carbon fiber keels have a property to store and return energy (Menard et al. 1992; Postema et al. 1997b; Nolan 2008).

2.5.2 Prosthetic sockets and suspensions

The purpose of the prosthetic socket is to provide an interface between the residual limb and the prosthesis. Amputees' residual limb supports the body through the suspension and the socket in bipedal locomotion, so a comfortable prosthetic socket has a critical role in an amputees' rehabilitation (Legro et al. 1999). The most advanced prosthetic device will not be in use if the socket (an interface) does not provide control. Currently, prosthetic socket material is plastic polymer laminate, urethanes, mineral-based liners, and improved silicones, are much more flexible and easily shaped (moulded) from a plaster-of-Paris cast of the residual limb to fit comfortably on the amputee (Gerschutz et al. 2011). The socket provides control of prosthetic device so discomfort could lead to the residual limb tissue injury (Chadderton 1978; Meulenbelt et al. 2007; Ebrahimzadeh and Hariri 2009). Residual limb tissue injury leads to pain and as a result to non-approval of a prosthetic device. In the long term study, Dillingham et al. indicated that less than a half of (43% from 146) patients were satisfied with their prosthesis comfort (Dillingham et al. 2001). Nevertheless, the design improvement of prosthetic sockets could have a positive effect on the biomechanics of gait. A vacuum socket fitting design demonstrates more symmetrical step length and stance compared to a total surface-bearing socket. The results have better mechanical and sensory control of the prosthetic foot due to good fitting and skin contact (Board et al. 2001). Nevertheless, even this should be interpreted carefully due to a deficiency in the methodology of the study as other components do not keep constant. Comfort should be considered the most important characteristic of the prosthesis (Legro et al. 1999) however prosthetic devices must possess two main characteristics: no pain in residual limb and no fatigue (Postema et al. 1997a; Postema et al. 1997b).

TTs currently utilise two main socket designs with combined modifications. First, the most typical is a conventional patellar tendon bearing (PTB) socket that was designed by the Biomechanics Laboratory of the University of California, in 1958. This socket employed pressure over 100 kPa: the patellar bar, the proximal popliteal area, the posterior medial flare and the fibula head (Convery and Buis 1998). The bearing is taken through the residual limb to a high socket that covers all the tendon below the patella to provide stability against side loads but allows optimal knee flexion/extension. This design is beneficial as it alleviates strain from sensitive areas of the residual limb such as the area between distal bone ends of tibia and fibula and proximal fibula head. The second socket design is a total surface bearing socket (TSB) which is contradictory to PTB postulation. The concept of TSB socket is to present a bond with the residual limb by evenly distributing pressure in all areas including sensitive ones used for weight bearing (Goh et al. 2003). The design of TSB sockets has been improving with the development of new gels that are used between the silicon liner and the socket. An additional benefit is that geometrical configuration only has a small margin of error in this type of interface. However, some modifications of TSB can include details from PTB as prime bearing would be applied on the patellar bar although another pressure bearing would be shared through all areas of the residual limb and socket. This could be used for immature or fluctuating residual limb (Figure 1) (Hachisuka et al. 1998). There are many prosthetic socket modifications within the two main designs to improve gait efficiency within the comfort of patients. Interestingly, an alternative technique of interface between residual limb and prosthetics employed skeletal attachment (*osseointegration*) that was implanted into a residual bone as an extension of an amputated part, which could improve amputees' gait (Eriksson and Branemark 1994; dos Santos et al. 2010). However, the use of this technique is very limited due to the risk of residual limb bone resorption and a high possibility of infection. To reduce the risk, the strut materials have to be improved with the maintenance of meticulous long-term hygiene (Nishimura et al. 1998).

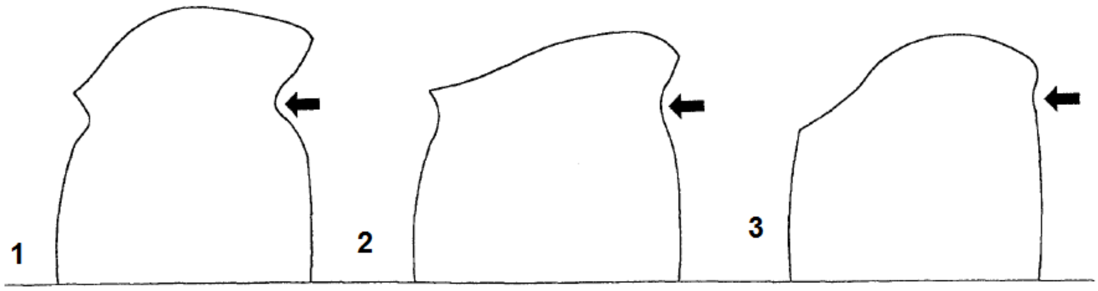

Figure 1. Prosthetic socket designs (1 - PTB socket; 2 – modified PTB socket; 3 – TSB socket). Arrows show where sockets are curved for patellar tendon. From Hachisuka et al. (1998) (Hachisuka et al. 1998).

The suspension sleeve is a durable junction and secure attachment between the residual limb and a prosthetic foot. A prosthetic suspension sleeve is typically made of neoprene, latex, urethane or other elasticised fabric. The suspension system for trans-tibial sockets is used with full contact sockets. The main suspension categories are vacuum (suction suspension), anatomical contour, strap and hinged. The first and the most typically used suspension employs a vacuum or simple suction function that depends on vale suction and silicon liner and creating a seal between the residuum skin and the prosthesis. This type of suspension is typically considered as a comfort for the amputees due to reduced pistoning. The downside of the suspension could be that the level of vacuum could decrease extremely during swing phase after many reiterations (Beil et al. 2002; Moo et al. 2009) as suction must be donned consistently. Full contact (TBS and hydrostatic) sockets are employed: vacuum (a one direction, rejecting valve for air out), additionally a sealing sleeve (neoprene, latex) that has two purposes: to keep air out and provide support to the socket from the residual thigh and anatomical (Supracondylar – Suprapatellar). The second category is an anatomical counter suspension employed when the suction suspension is not possible. This category was first used in the PTB socket with the name Supracondylar – Suprapatellar which is specially designed to stabilise the knee and suspend the prosthesis by improving the socket over the condyle and patella. Hence, make femoral condyle, and patellar of the residual knee are completely in the socket. The system provides medial-lateral stability with stabilising the residual knee and preventing varus/valgus. Additionally, amputees for Supracondylar – Suprapatellar suspension are wearing a thigh residual limb corset (Rubin et al. 1970). The tubular sleeve could be worn in conjunction with a vacuum and anatomical counter suspensions over the socket to lengthen the patent residual limb and create a secure link between the socket and the residual limb. The sleeve can be manufactured from neoprene, urethane, latex or other similar materials. The third category is strap suspension that combines various bands to secure the socket and could be used in combination with vacuum suspension. The fourth, historical suspension, is least common a hinge suspension also can be named 'joints and corset' where is used the thigh Lacer with the hinge transmitting suspension. To secure a link between the socket and the residuum patients could use more than one type of suspension. The choice of suspension system depends on the requirements of patients and is affected by various factors such as anatomical contours, activity level, weight and inertial properties of the prosthesis, personal preferences and even geographical location.

The connector between the socket and prosthetic foot is a pylon which additionally could provide a shock absorption function (Klute et al. 2001). Shock-absorbing pylons reduce the vertical shock, store and partly returns this energy during a gait cycle that improves comfort and their efficiency (Berge et al. 2004).

One of the main considerations to deliver an effective rehabilitation process is to have a comfortable socket. Prosthetists select prosthetic socket types according to the anthropometry of the patient and their experience. The most commonly, used prosthetic socket is a patellar tendon bearing (PTB) however, some modification could include a total surface bearing (TSB) design. To ensure a comfortable fit between the prosthetic socket and the patient's residual limb a suspension sleeve is used, which is selected according to the socket used.

Finally, the connector between the prosthetic socket and prosthetic foot is a pylon which commonly includes the shock absorption function.

2.5.3 Prosthetic ankle-foot devices mass consideration

Development of the prosthetic ankle-foot device is focused to replicate the functionality of a healthy biological human foot to perform a wide range of physical activities. However, patients have different weight, gait pattern, levels of activity and needs. Hence, prosthetic manufactures would benefit from a modulated design of prosthetic devices. Where the estimation of optimal prosthetic mass with its inertial properties is dependent on several variables and is considered a challenging task. Those variables are patient activity level, residual limb length, muscle volume in the residual limb, knee muscle strength of flexors/extensors and other patient personal considerations. Early studies suggested, the lower-limb prosthetic device has to be as light as possible to minimise muscle work and energy expenditure during locomotion (Ralston and Lukin 1969; Godfrey et al. 1977) because heavier prosthesis requires more work for the initiation (acceleration) and termination (deceleration) to perform a stride. Hence, the key consideration for designing prostheses was a lighter mass. Conventional prosthetic devices typically 30-40% lighter then intact side (Lehmann et al. 1998) where TTs have shown asymmetrical gait pattern (Lehmann et al. 1993; Sanderson and Martin 1997). Several studies investigating effects of added mass (up to 100% of the estimated intact limb below knee mass) on gait symmetry, however, the use of such a prosthesis led to increasing of gait asymmetry and metabolic cost compared to prosthesis without added mass (Lehmann et al. 1998; Mattes et al. 2000; Smith and Martin 2013). Despite this, another research has shown that heavier prosthetic devices assist the propulsion of the trunk forward (Gitter et al. 1997; Lehmann et al. 1998) without notably increasing metabolic cost in TTs (Gailey et al. 1997; Lehmann et al. 1998; Lin-Chan et al. 2003). A possible benefit of a heavier prosthesis could include the maintenance of balance during amputees' locomotion. Prosthetic lower-limb designs have evolved, so an adaptive prosthesis functionality has shown improvements in the biomechanics during different activities (Alimusaj et al. 2009; Wolf et al. 2009; Fradet et al. 2010). Improvement of lower-limb prosthetic devices functionality commonly leads to an increase in its mass. Heavier lower-limb prosthetic devices with improved functionality have demonstrated a reduction in metabolic cost in trans-femoral (Buckley et al. 1997) and trans-tibial (Au et al. 2009) amputees.

However, gait symmetry depends not only on prosthetic mass but also on this mass distribution where the Centre-of-Mass location is relative to the axis of oscillation affects inertial properties. The proximal Centre-of-Mass location has presented a decrease of prosthetic limb swing time which more closely matches the intact limb of TTs (Mattes et al. 2000). Hence, the proximal Centre-of-Mass location is considered as being more optimal in the prosthesis for self-selected walking speed as a result of examination of the energy cost and gait pattern of amputees for gait efficiency (Tashman et al. 1985; Mattes et al. 2000; Smith and Martin 2013). To estimate a prosthetic moment of inertia that is unattached to the patient an oscillation technique was used, but making calculations throughout the amputees' gait requires different approaches. The calculation of the prosthetic limb inertial properties is a result of combined evaluation of a residual limb volume and Centre-of-Gravity location of the prosthesis (Miller 1987). This calculation has been employed by a number of researchers (Winter and Sienko 1988; Buckley 2000). To calculate moments of inertia *Newton's First Law of Motion* can be used. The equation below represents representing angular inertia about the centre of gravity.

$I_0=mk^2$ Equation 1 (Gordon et al. 2004)

Where m is the mass of the segment and k is the radius of gyration.

Parallel axis theorem below is shown the moment of inertia of a segment about any arbitrary axis (Gordon et al. 2004).

$$I_k=I_0+m_s r_1^2+m_w r_2^2 \qquad \text{Equation 2}$$

In Equation 2 I_k - a moment of inertia for a segment of prosthesis or leg below the knee, m_s and m_w weight of segment and additional mass respectively, r_1 and r_2 distance from axes of rotation till Centre-of-Mass segment and extra weight respectively. For amputees, m_s includes the mass of the prosthesis plus the mass of residuum.

Typically, improvement of prosthetic device functionality leads to an increase of device mass. Certainly, enhanced prosthetic functionality would benefit amputees' biomechanics of gait. The mass of the prosthetic device would also have an effect on the biomechanics of gait. This effect is mostly during the swing phase, so the device CoM location is also critical. Perfect prosthesis mass with CoM location should be able to provide as close as possible metabolic energy cost with optimal gait pattern as able-bodied individuals. However, there is currently is no known lower-limb prosthetic device that can deliver this. Although, the literature presented that the functionality of prosthetic devices improves gait, so an insignificant increase of prosthesis mass does not have the negative effect on gait efficiency. Little is known about the effect of lower-limb prosthesis mass during the stance phase. Indeed, the prosthesis mass distribution would likely have no direct effect during stance phase, but it could be a consequence of swing phase. The studies mentioned above have investigated overground gait, but little is known about the effects of the heavier prosthetic device with improved functionality during ramp descent. Where gravitational potential energy has increased compared to overground gait.

2.5.4 Prosthetic ankle-foot design development

Currently, lower-limb amputees use two main categories of prosthetic ankle-foot devices: rigid and articulated (Edelstein 1988). The most frequently used is a rigid ankle-foot device such as Solid Ankle Cushioned Heel (SACH) feet or dynamic response feet. SACH often prescribed to amputees with limited activity level, ability, and weight (Hofstad et al. 2004; Marinakis 2004). The design of a dynamic-response foot has contributed more response compared to other designs by increased ability to absorb, store and release more energy. The benefits of dynamic response feet have been presented in many studies (Edelstein 1988; Alaranta et al. 1994). Initially the dynamic-response foot was designed for active amputees; however, despite this even less active amputees find the design helpful. Improved design of dynamic response feet has an integrated articulated ‘ankle’ mechanism. The research of Su et al. (2010) has presented that walking down slopes perceived to be easier with such a prosthesis (Su et al. 2010).

To perform safe ramp descent required an increase of control of the body weight forward/downward transition compared to overground gait (Smith et al. 1998). To control the increased potential gravitational energy (Chapman 2008) the ankle is required to plantar-flex until foot-flat. To attain foot-flat quicker is crucial for anterior-posterior stability on the ramp (Redfern et al. 2001). The body weight fall from the contralateral limb have increased for ramp descent compared to overground gait, so the knee flexion loading response also increased (Lay et al. 2006; Lay et al. 2007). Following attainment of the foot-flat, the ankle dorsi-flexors has to control the forward rotation of the tibia over the foot to control increased potential gravitational energy during ramp descent (Lay et al. 2006; Lay et al. 2007). Then push-off requirements at the ankle reduced because body weight transition assisted by increased potential gravitational energy (Lay et al. 2007). Consequently, the use of the prosthetic ankle-foot device during ramp descent would lead to adaptations in remaining joints of the lower-limb system within changes of the locomotor function.

The functionality of the prosthetic foot device with rigid ‘ankle’ is based on the properties of the heel and fore-foot keel materials, where the heel and fore-foot keels have constant stiffness properties. The geometrical configuration and stiffness of the heel and fore-foot keel depends on the required prosthetic functionality, patient weight and gait characteristics. During the initial contact, the prosthetic heel keel has to provide shock absorption to reduce the impact on the residuum and whole body through the socket. The function of the heel keel is to provide power absorption or braking during initial contact and depends on the properties of the material used. Heel absorption followed by an imitation of plantar-flexion to achieve foot-flat and provide weight-bearing stability. Prosthetic foot designs have to consider stiffness of the heel keel spring during initial contact as there is the impact on external knee extension moment, so the hamstring muscles maintain the knee in a flexed position. During slope descent, rigid ‘ankle’ would not be able to attain foot-flat quicker as heel keel stiffness has been selected for overground gait, so to attain foot-flat quicker, the pylon/shank have to rotate forward after initial contact. Rotation of the pylon/shank would lead to an increase of loading response knee flexion. The following single-limb-support phase is not directly affected by the stiffness of the heel or fore-foot keel springs directly. However, the effect could be as a response from heel keel after initial contact. The single-limb-support progression and pre-swing phase of the prosthetic device, the keel stiffness contributes towards progression (Perry et al. 1992). During slope descent, this contribution towards progression could have an adverse effect due to increased potential gravitational energy (Chapman 2008) that assists the body weight transition over the support limb, as safe slope descent require control of the body weight transition. In overground gait, the prosthetic foot device functionality in the following phase require to assist body progression with the necessary momentum to roll over the contralateral foot, so fore-foot keel stiffness during the late stance phase is required to return energy before the swing phase. The energy started storing during mid-stance and realised when the body weight starts transferring to the contralateral side. The fore-foot keel responds by adding energy for the limb to swing and to propulse the body forward. During slope descent, the importance of the fore-foot keel ‘push-off’ to propulse the body forward from the fore-foot keel reduced, because potential gravitational energy contributes into the body weight transition. Typically, the design of a dynamic response foot has a split fore-foot keel that provides eversion/inversion with the foot’s ability to roll from side-to-side on uneven terrain without losing balance or energy return – to replicate the intact foot.

Articulated ankle-foot prosthetic devices are commonly designed in cooperation with a dynamic-response foot to add motion to the prosthetic device. Use of multi-axial prosthetic ankle-foot devices on uneven surfaces reduces stress between socket and residuum by more absorption compared to a rigid ankle-foot device. Articulated ankle-foot designs included elastic bumpers (rubber - snubber) or with visco-elastic dampers (hydraulic) mechanisms. The use of hydraulically damped mechanism may improve comfort and protect from the damage caused to the soft tissue of the residuum from high stresses due to a reduction in in-socket pressure in TTs (Portnoy et al. 2012). The use of hydraulically articulated ‘ankle’ attenuated the disruptions in Centre of Pressure progression (De Asha et al. 2013a) with increased self-selected walking speed (De Asha et al. 2013b). The prosthetic ‘ankle’ articulation mechanism typically has a biomimetic location of the intact ankle and attached at the end of pylon to tripod construction (the heel and fore-foot keels). The users of the various hind-foot rollers have an effect on shock absorption, weight-bearing stability, and progression. The study of Su et al. presented that downslope gait is observed to be easier for users of articulated ankle-foot devices (Su et al. 2010). Nevertheless, non-adaptable ankle-foot articulation designs have also been set for overground walking and self-selected walking speed. Consequently, change a walking speed, approaching stairs, ambulating inclined surfaces with the non adaptable ankle-foot device could have a negative impact on the biomechanics of gait.

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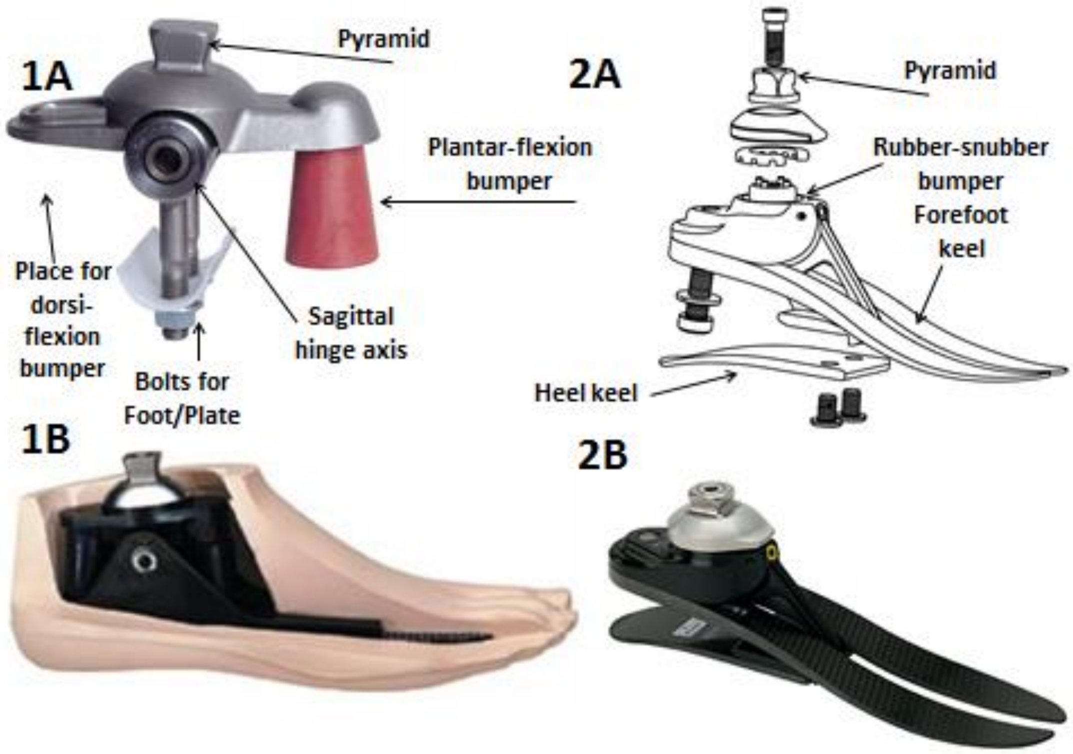

Figure 2. Prosthetic foot designs. 1A and B - Single-axis ankle-foot device schematic (top) and photograph (bottom) (Ohio Willow Wood Co. (Mt. Sterling, OH) adapted from www.willowwoodco.com; 2 A and B – Multi-axis ankle-foot device *Epirus* schematic(top) and photograph (bottom) (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) which is used in this thesis. Adapted from www.blatchford.co.uk. Accessed 11.05.2016.

The multi-axial ankle-foot prosthesis can be separated into the fore-foot and hind foot designs. Hind foot designs utilise elastic (rubber, snubber) or visco-elastic (hydraulic) properties and its geometry (spherical, ring) for the articulation of an ankle-foot device. From a simple hind-foot articulation there is a 'rubber-snubber', (*Epirus*; Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) to a more advanced hydraulic dampening (*Echelon*; Avalon^{k2}; Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). The main benefit of an elastic hind foot articulation is it is lightweight compared to visco-elastic. However, visco-elastic hind foot articulation has the disadvantage of heavier weight with a more qualified service, but as mentioned earlier, an optimal weight of the prosthetic device and optimal weight distribution has not been estimated yet. The ankle-foot prosthesis with visco-elastic hind foot articulation has biomechanical advantages in unilateral TTs (De Asha et al. 2013b; De Asha et al. 2014). The mechanism of a hind-foot roller is mounted in an ankle-foot prosthetic device approximately in the location of the biological ankle. This mechanism of the hind-foot roller responds to load according to the stance phase and its transition. The hind-foot roller could indirectly contribute towards shock absorption and with optimal set up provide the right timing for a keel to perform push off. The main design of fore-foot multi-axial ankle-foot is a split toe that helps to provide stability, particularly on uneven surfaces. Hindfoot simulates 'plantar-flexion' with inversion/eversion response during loading response to adapt to the approached surface. In multi-axial hind foot designs, hind foot increase translational motion that extends, providing fore-foot or heel of prosthetic ankle-foot response.

Figure 3. Schematic (top) and photograph (bottom) showing the microprocessor control quasi-passive hydraulic ankle-foot device (*Elan*; Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) which is the subject of this thesis. Adapted from www.blatchford. Accessed 15.05.2016.

The development of dynamic-response prosthetic feet and hind-foot rollers required an update of the classification used. In the category of articulated ankle-foot designs can be added adaptable ankle-foot prosthesis that could change plantar/dorsi-flexion resistance according to the slope of ambulation.

The advanced design of the hind foot is an adaptive mechanical device that has improved ankle-foot functionality through the use of a microprocessor. An adaptable microprocessor controlled hydraulic quasi-passive prosthetic ankle-foot device (*Élan*; Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) was designed to adapt to different terrains, walking speeds. The adaptation operated by microprocessor controlled hydraulic ‘ankle’ by increasing or reducing plantar/dorsi-flexion resistance to achieve smoother, safer and more natural gait pattern. For example, safe slope descent requires control of body weight motion due to increased potential gravitational energy, so it is critical to establish a foot-flat sooner and have controlled transition over the support foot. The manufacturer claimed, that microprocessor controlled hydraulic ankle-foot device reduce damping to simulate ‘plantar-flexion’ to attain foot-flat sooner. A subsequent increase of ‘dorsi-flexion’ resistance should delver control of body weight transition over the support foot. Nevertheless, this claim has not yet been supported by independent research. The analysis of amputees response on different prosthetics while performing different tasks should be accessed in order to improve prosthetics design and rehabilitation process.

The manufacturer claimed that the *Élan* ankle-foot device adapts to the user walking speed, by providing a maximum return, when necessary from e-Carbon spring stored energy. The device's microprocessor should respond to the user’s increase of walking speed by the increase of ‘plantar-flexion’ and decrease ‘dorsi-flexion’ resistances. The manufacturer also claimed that the *Élan* increases body propulsion forward (maybe due to weight). The *Élan* device is also claimed to adapt slopes. The device eases slope ascent by increasing ‘dorsi-flexion’ resistance and provide safe and controlled slope descent by reducing ‘dorsi-flexion’ resistance until foot-flat which follow by increasing ‘plantar-flexion’ resistance.

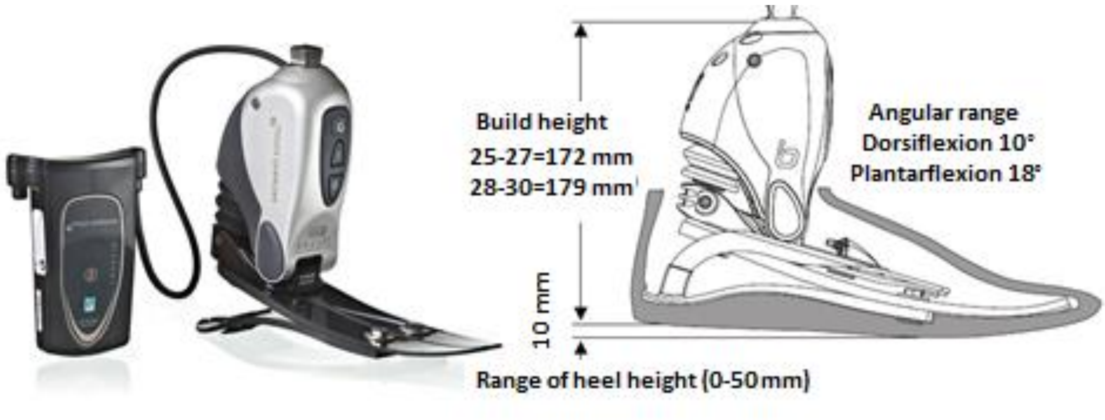

Figure 4. Proprio-Foot photograph (left) and schematic (right) (Ossur, Reykjavik, Iceland). Adapted from www.ossur.co.uk. Accessed 20.05.2016.

The analogous adaptable device is *Proprio-Foot* from Ossur (Ossur, hf, Iceland) (Figure 4). However, there is a difference in mechanical functionality, the adaptation of powered ankle-foot device *Proprio-Foot* (Ossur, hf, Iceland) occurs only in the swing phase, but in stance phase acts as the conventional dynamic-response foot that was indicated by a number of researches (Versluys et al. 2008; Eilenberg et al. 2010). Previous investigations of the adaptive foot *Proprio-Foot* (Ossur hf, Iceland) questioned its benefits during slope descent, due to a less physiological gait (Fradet et al. 2010). The researcher proposed that the effect of the ankle-foot device could be more visible on a higher gradient, but a 7.5º gradient is notably steeper than the maximum suggested disabled ramp gradient 5º (Alderson 2010). Nevertheless, the patients' of the study indicated feeling safer and had reduced stress on the knee joint (Fradet et al. 2010). Certainly, there are other hydraulic ankle-foot prosthetic devices with adaptive functionality such as Raize-foot (Fillauer, USA), Meridium (Otto Bock, Germany), Triton Smart Ankle (TSA) (Otto Bock, Germany). However, there were no current scientific publications as the performance data is unavailable for further analysis. Therefore, the above-mentioned devices have been excluded in further review.

The sections have described and discussed the ankle-foot design, development of various prosthetic foot devices which are used by TTs gait. Conventional prosthetic foot designs have set up for overground gait with self-selected walking speed, so the slope descent with such device could have a negative impact on the biomechanics of gait. Detailed examination biomechanics of slope descent in trans-tibial amputee patients with various ankle-foot articulations. This would allow a direct comparison between ankle-foot articulations. The analysis and patients' feedback will provide deeper understanding biomechanics of gait with assessed prosthetic ankle-foot components. The following section describes and discusses the gait cycle.

2.6 Gait cycle

The current 3D motion capture systems and force plates allow detailed analyses of human movement. Gait analysis focuses on the lower-limb. The function of the lower-limb is to support body weight against gravity with propulsion forward during gait (Sadeghi et al. 1997; Sadeghi et al. 2001). The assessment of biomechanics during different gait tasks in lower extremity amputees could help provide a better understanding of body locomotion within the development of prosthetic devices. The process of assessment of cyclic body locomotion is termed: gait analysis where the assessment of joint kinetics employed inverse dynamics. The gait cycle is divided into stance and swing phases during cyclic body locomotion and illustrated in figure 5. The gait cycle can be identified from initial foot contact to the next subsequent foot contact with the same foot. Throughout the gait cycle, feet attained double- support and single-limb-support. Double limb support where both feet are in contact with the ground which takes place at initial and terminal stance phase. Double limb support time reduces with increased walking velocity. The gait of healthy individuals with customary speed, initial and terminal double limb support have around 12% of the gait cycle each with total double limb support 25% of the gait cycle (Ayyappa 1997). The single-limb-support (SLS) where one foot is in contact with the ground and another in swing phase. The stance phase can be divided into three functional rockers (Perry et al. 1992). Three functional rockers: Weight Acceptance, Single-limb-support, and Limb Advancement presented. The stance phase can be presented in five phases: initial contact, loading response, mid-stance, terminal stance and pre-swing (Figure 5). The following swing phase can be presented in three phases: initial swing, mid-swing, terminal swing. The focus of this thesis is stance phase, so further analysis of the lower extremity motion will assess the five stance phases in three functional rockers. Understanding the effects of that rocker on gait biomechanics will help to evaluate the effect of orthotics and prosthetics on gait efficiency. These three functional rockers are described as follows.

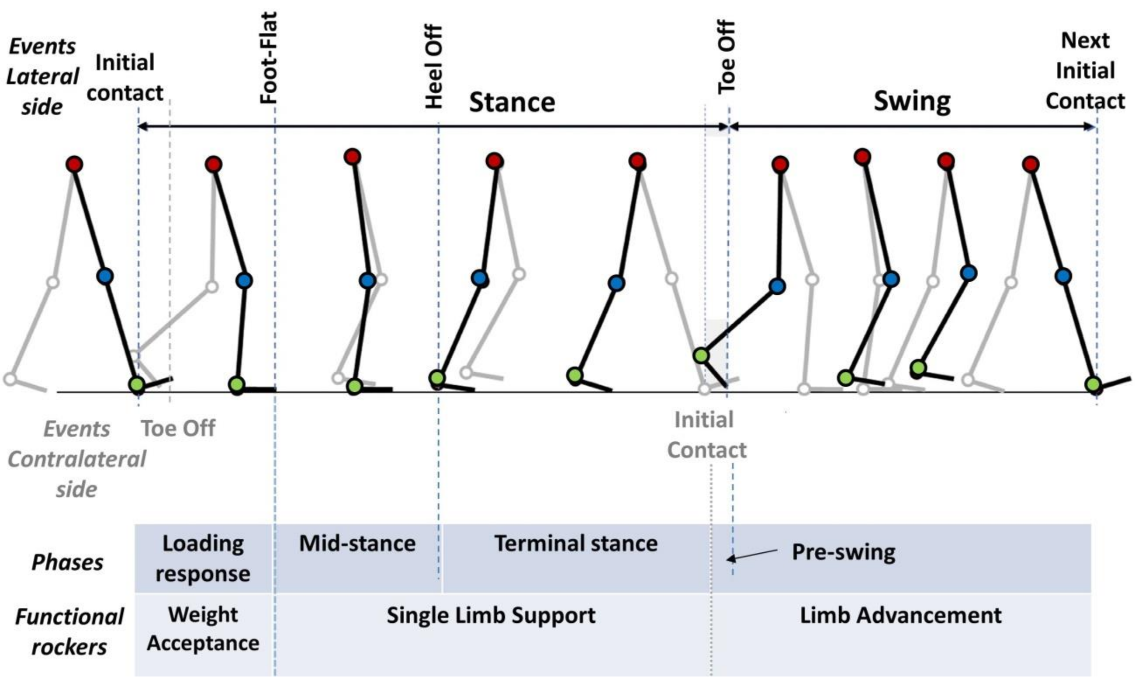

Figure 5. Schematic representation of lower-limbs kinematics in able-bodied individuals. The gait cycle with functional rockers, phases and events highlighted. Adapted from Perry et al. (1992) (Perry et al. 1992).

Weight Acceptance, Heel rocker, this could also be referred to as the First rocker (Figure 5). During the initial double support phase, the momentum generated by the fall of the body weight transfers to the lead limb from the trail. The stance phase begins from the initial foot contact and progression to foot-flat that is achieved by the ankle planter-flexion motion. Initial contact is commonly attained by the calcaneal tuberosity with fulcrum motion between the foot and tibia. The foot motion via the fulcrum preserves forward progression. Therefore, the term Heel contact or Heel strike was used to describe the initial contact with the ground. However, this event would be more appropriate to term as an initial contact because the heel velocity immediately prior contact with the ground is almost zero vertically and low (~5% of maximal heel velocity) in the horizontal direction (Winter 1992). The rocker ends at the contralateral foot toe off from the ground with the subsequent following to SLS. During heel rocker, the knee flexion angle changes from an extended position to a more flexed position to absorb shock with body weight which is fully transferred into the lead limb which is termed as a loading response (Kirtley 2006a). The loading response could be presented by the first vertical ground reaction forces (vGRF) peak. The function of the rocker is to translate the vGRF into the forward progression of the shank with initial limb stability. The magnitude of vGRF during heel rocker is typically over one body weight of the individual. Anterior-posterior ground reaction forces (A-P GRF) indicating braking force during the first rocker. The medial-lateral ground reaction forces (M-L GRF) typically increasing in a medial direction, however, the magnitude of this force is not significant and depends on the gait of an individual (Kirtley 2006a). During the initial double support the limb delivered by the moments of ankle plantar-flexion, knee flexor and hip extensor. However, ankle joint after initial contact has a short period of dorsi-flexor muscle activity that helps to control lowering foot-flat to the ground which occurs during the first 10% of the gait cycle (Winter et al. 1995). There is an active ankle plantar-flexor moment involved with up to 50% of the gait cycle (Czerniecki 1988). During the initial contact minor power is absorbed by the ankle, but major occurs in the knee joint power. The significant knee joint power is the result of the knee flexion, which is controlled by the eccentric knee extensors. Several publications have documented that the hip extensors are acting concentrically to deliver power generation for forwarding progression (Sadeghi 2000; Kirtley 2006b). Hip power generation occurs after initial heel contact followed by controlled forward motion of the trunk (Winter et al. 1995) and controlled collapse of the support limb (Sadeghi 2000). Therefore, the hip extensors power prevents lower-limb collapse and stabilises the trunk (Perry et al. 1992; Eng and Winter 1995). With the support of the BW before foot-flat (Neptune et al. 2004) due to reducing BW support on a contralateral limb prior pre-swing phase (Winter 1991; Perry et al. 1992).

Single-limb-support, Ankle rocker, and could also be referred to as the second rocker (Figure 5). During the second rocker, the tibia progresses in an arc trajectory over the support foot. The second rocker is defined throughout the swing phase of the contralateral foot and is associated from mid-stance through terminal stance. Mid-stance is the first half of the single-limb-support and begins at the contralateral foot off the ground and continues until body weight is aligned over over the fore-foot. During mid-stance, the support limb shank rotates over the foot with ankle joint motion from a plantar-flexion to a dorsi-flexion position. The function of the rocker is to provide stability with control of forward body velocity as the shank rotates over the support limb with foot on the ground at the ankle joint (Perry et al. 1992). During mid-stance phase, vGRF magnitude is dropped from above body weight (around 1.2 – 1.3) to below body weight (around 0.7-0.8). In the mid-stance phase, this involves hip flexor moment with power absorption which enables body weight to progress to a terminal stance. In the terminal stance phase, the swing of the contralateral limb is reducing vGRF from support limb with change a-pGRF from braking to propulsion (Kirtley 2006a). Throughout the second rocker, the ankle joint has an active plantar-flexor moment (Czerniecki 1988) to control shank forward rotation (Winter et al. 1995). The function of the ankle could be presented as a fulcrum in the inverted pendulum model. The knee joint moments are transferred from extensor throughout to flexor manner. Mid-stance is the period immediately following loading response; the knee begins to extend and provide power generation to achieves the leg near to full extension, which reduces the fall of the pelvis at contralateral foot contact. Contralateral limb in the swing phase, the support leg, knee extends to ensure safe swing (Kirtley 2006c). During mid-stance, to support body weight within balance is active hip flexors. The second half of the single-limb-support is termed as a terminal stance. The subsequent motion of tibia after mid-stance follows to terminal stance and continues to dorsiflex ankle joint and extend the knee to prepare for the pre-swing phase, where the hip joint is flexed but continues to extend. The phase begins when the support limb heel rises or heel off and finishes when the contralateral foot contacts the ground (Perry et al. 1992). The ascending second peak vGRF presents terminal stance.

Limb Advancement, Fore-foot rocker, also referred to as the Third rocker. This starts from the contralateral foot contact till toe off (end of stance) or the beginning of swing phase (Figure 5). Another suggestion of limb advancement begins when the limb is swinging to the contralateral (Perry et al. 1992). However, in gait pathologies with the inability to lift a foot from the ground the swing phase the term foot drag can be used. There are reasonable explanations of ankle plantar-flexion function during the third rocker; to propel the body upward/forward (Winter 1983); to restrict the trunk over the ankle and to assist motion of the limb into the swing phase (Inman 1966; Cappozzo et al. 1976); with a small contribution to maintain CoM height against gravity (Meinders et al. 1998). During the rocker, foot contact with the ground has a further influence on prolonging the swing phase. The heel of the support limb rises with the fulcrum for tibial advancement transfers forwards to the metatarsal heads, propelling body weight forward. This is produced by an ankle plantar-flexor moment with simultaneous power generation by the triceps surae. The terminal double-limb support is a pre-swing phase that contains around 12% of the stance phase. The phase starts at approximately 50-60% of the gait cycle with the pre-swing phase and finishes by the end of the stance phase. The function of the phase is a safe setup transition from terminal double support throughout swing phase. It begins when the contra-lateral foot contacts the ground and ends with an ipsilateral toe-off. During this period, the stance limb is unloaded, and body weight transferred onto the contra-lateral limb. The descending portion of the second peak of vGRF demonstrates the period of pre-swing phase. During limb advancement phase the limb is getting ready for a swing. The ankle joint is plantar-flexed with the concentric power to provide propulsion of the limb forward into swing phase. The hip changes from extension to flexion together with flexion of the knee to ensure safe foot clearance during swing phase (Kirtley 2006c).

Throughout the following *swing phase,* the foot of the swung limb is not in contact with the ground and contralateral in stance phase. The swing leg acts as a compound pendulum, where the period controlled by the mass and centre of mass location (Tashman et al. 1985; Perry et al. 1992). The phase is divided on the initial swing, mid-swing, terminal swing. The initial swing (62 to 75% of the gait cycle) began at toe off and ends when the swinging limb is aligned with the contralateral limb. The foot from plantar-flexion (push-off) is changing to a dorsi-flexion position in order to achieve adequate foot clearance. The lift of the foot from the ground is also lead to an increase in the knee and hip flexion. The next phase is mid-swing which begins when the swinging limb is aligned with the contralateral limb until shank of swing limb is in a vertical position. The phase occurs for the period from 73 to 87% of the gait cycle (Perry et al. 1992). For a period of mid-swing, the swung leg knee flexion still has an important role for adequate foot clearance. During mid-swing phase, the event minimum foot clearance is the minimum vertical distance between the ground and the toes region that occurs around the instant when the foot travels with maximum horizontal velocity (Winter 1992). At the point of minimum foot, clearance is the highest risk of tripping that could lead to falls (Blake et al. 1988; Mills and Barrett 2001). The examination of minimum foot clearance in healthy individuals (group mean) has been reported for young adults 1.29 cm and elderly 1.12 cm (Winter 1992; Karst et al. 1999). The mid-swing phase. The final phase of the gait cycle is terminal swing (85 to 100% of the gait cycle) with full extension of the knee before initial foot contact (Perry et al. 1992). The ankle joint is maintained in a comparatively neutral angle position throughout the initial swing phase (Kirtley 2006c).

The human musculoskeletal system is bipedal and functions to provide efficient locomotion (Lovejoy 2005). The sinusoidal curve of a Centre-of-Mass motion in the vertical direction has a displacement of 3-4 cm (Saunders et al. 1953). During the non-pathological gait cycle, the highest point of Centre-of-Mass is the single-limb-support, and lowest is double limb support. At initial double support, gravitational potential energy rises to a Centre-of-Mass highest point and returns as kinetic energy with Centre-of-Mass fall after the highest point, which passively utilises an inverted pendulum model (Cavagna and Margaria 1966; Usherwood et al. 2008). The efficiency of the gait depends on kinetic energy recovery. At the most efficient self-selected walking speed (Figueiredo et al. 2013) up to 65% is recovered from the energy saving mechanism

(Cavagna et al. 1977). Remaining energy has to be contributed by the muscles. Hence, the application of an inverted pendulum theory in locomotion could explain lessened muscle work relative to total energy consumption.

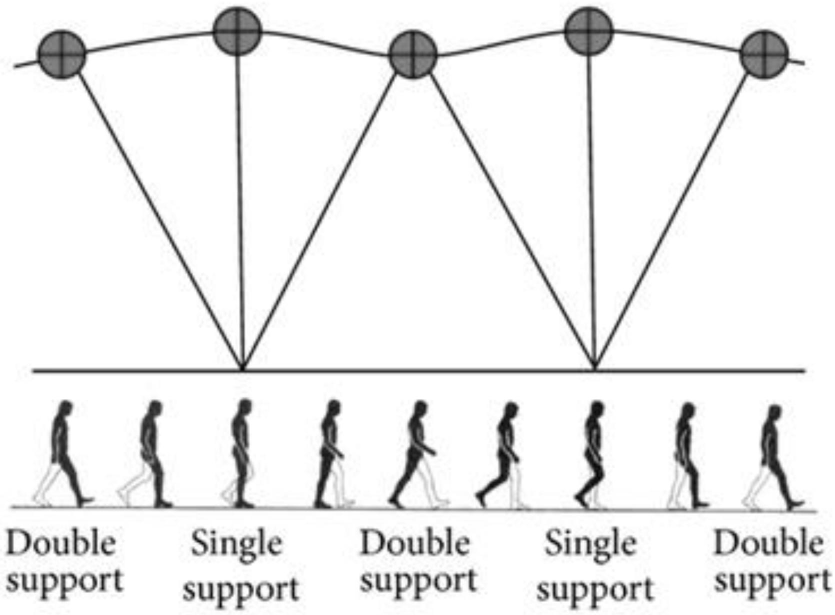

Figure 6. Inverted pendulum model of gait as the single-limb-support motion over the supporting foot in three rockers. Adapted from Lobet et al. (2013) (Lobet et al. 2013).

The bipedal locomotion of human gait can be described by using a number of prevailing theories. Human locomotion is exceptionally coordinated by complex interactions between limbs and all the segments to provide the smooth functioning of the whole within the neuromuscular system (Hunt and McPoil 1995). Physical loss or elimination of motion of one segment will affect the functionality of the whole system with compensation by other parts (Winter 1990). The function of the human biomechanical system is to provide efficient and safe locomotion across different terrains with various speeds. Inverted pendulum model describes motion between absorption and propulsion delivered by muscles within its conjoint cooperation. However, the theory has not been fully investigated with interferences in ankle motion. Investigation of ankle motion/function is critical as it plays the role of a fulcrum in the inverted pendulum model (Kuo 2007).

2.7 Overground gait, able-bodied individuals versus unilateral trans-tibial amputees

2.7.1 Spatio-temporal parameters

The spatial-temporal parameters are most commonly recognised as a clinical assessment of gait pathology. This assessment is frequently used to evaluate the symmetry of gait (Nolan et al. 2003). The spatial-temporal parameters included the variables such as time (stance time, swing time, time to attain foot-flat.), the distance metrics (stride length, step length and stride width) and the variables that linked time and distance (walking speed, cadence). The temporal parameters could also examine single and double-limb support and foot-flat, according to the percentage of stance. The illustration of spatio-temporal parameters is presented in figure 7.

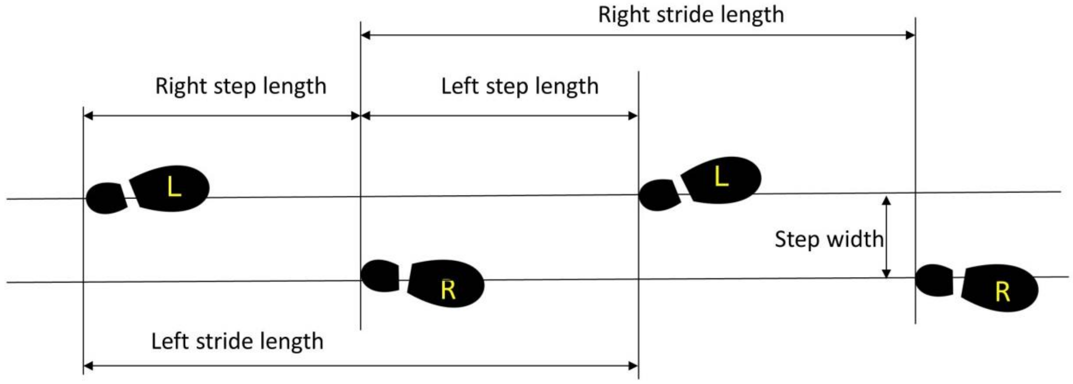

Figure 7. Spatio-temporal parameters to assess the symmetry of gait: step length, stride length, step width.

The ankle mechanism plays a critical role in achieving a human's safe and efficient bipedal locomotion. Ankle function contributes to an optimal gait pattern to transfer CoM with minimal metabolic energy cost (McNeill A 2002). Walking speed is a basic assessment of gait quality in individuals with gait disorder or lower extremity amputees. As reported by Bohannon (1997), who assessed healthy men and women in 20s, 30s, 40s, 50s, 60s and 70s years old, mean self-selected walking speed ranged from 1.27 m/s to 1.46 m/s (Bohannon

1997). However, prosthetic feet do not have the functionality of the biological feet, and as a result lower-limb amputees walk more slowly (Waters et al. 1976; Sulzle et al. 1978; Boonstra et al. 1993) with a significant increase of energy consumption compared to individuals without limb impairment (Gonzalez et al. 1974; Waters et al. 1976; Fisher and Gullickson 1978). Various studies confirmed that self-selected walking speed of TTs from 1.04 m/s to 1.11 m/s is slower than able-bodied individuals (Robinson et al. 1977; Kegel et al. 1981; Colborne et al. 1992). Nevertheless, self-selected walking speed depends on the prosthesis design, the study results of Nielson et al. (1989) presented, that TTs with dynamic response feet compared to SACH having significantly higher walking speeds (Nielsen et al. 1988). De Asha's comparative study (2013) found that TTs with dynamic response feet with hydraulically articulated ankle-foot device compared to dynamic response feet with the right ‘ankle’ have significantly higher walking speeds (De Asha et al. 2013b). Furthermore, walking velocity can be achieved by manipulation of step length (Figure 7) and step frequency (cadence) within biomechanical limitations of the musculoskeletal system (Nilsson and Thorstensson 1987). As reported by Kirtley (2006), able-bodied men during self-selected walking speed have a stride length 1.4-1.6m and step frequency (cadence) approximately 110-115 steps/min (Kirtley, 2006). Therefore, reduction of walking velocity of TTs compared to individuals without limb impairment could be the result of reduced stride length (Skinner and Effeney 1985; Barth et al. 1992). There, the use of dynamic response foot (Flex-Foot) (1.35 ±0.19 m) feet compared to SACH (1.25 ±0. 16 m) prosthetic feet have increased the stride length (Lemaire et al. 1993). These findings further support, that the use of a dynamic response foot would have a positive effect on spatio-temporal parameters of the amputee gait. Previous studies have reported, that lower-limb amputees have presented reduced stance time (Breakey 1976; Murray et al. 1983) and reduce load (Engsberg et al. 1991; Engsberg et al. 1993) on the prosthetic compared to intact limb. Lower-limb amputees spend less time on the prosthetic limb with reduction of single-limb-support could be the result of discomfort, pain, or absence of confidence in the prosthetic limb (Nolan et al. 2003). In a different study, Highsmithet al. (2010) has reported that step time correlated to stance time for both limbs in TTs (Highsmith et al. 2010). Greater stride time in TTs (prosthetic 1.160 s; intact 1.166 s) than able-bodied (1.065 s) indicated slower walking speed (Kendell et al. 2010). There, self-selected step frequency has chosen to minimise metabolic energy cost (Cotes and Meade 1960; Workman and Armstrong 1986).

The main goal of amputees' rehabilitation process is to restore an optimal gait pattern to the patient. The term, gait symmetry, is when both limbs behave identically without statistical differences between biomechanical parameters that measured bilaterally (Griffin et al. 1995; Gabbard 1997). A basic methodology in the evaluation of gait symmetry is the spatio-temporal ratio between right and the left, step length or stride cadence (stride/min). Healthy individuals without a gait disorder have a ratio around 1, but an increase or reduce of the ratio between values of the left and right limbs indicate gait asymmetry. A certain level of asymmetry is expected, even in healthy individuals without the gait disorder. To define an acceptable level of asymmetry, the 95% confidence interval was calculated for each measured parameter for healthy individuals without the gait disorder. If the calculated value fell outside the 95% that parameter would be considered as asymmetrical (Patterson et al. 2010). For example, the study presented stance time (1.02 ±0.02 s) with upper confidence interval boundary (1.06 s) over 81 healthy individuals (Patterson et al. 2010). An alternative study presented, that gait asymmetry in healthy individuals could be ± 4% of lower limit and ±13.00 % of the upper limit (Herzog et al., 1989). A different study in addition to work of Herzog et al. (1989), Knutson (2005) provides that 90% of healthy individuals have lower-limb anatomic inequality (Knutson 2005) which could lead to gait asymmetry. Clinicians, generally use basic spatio-temporal parameters as step length, cycle length and stance time between limbs (Breakey 1976). The symmetry of gait has some reasons for realisation. The first is visual or aesthetical as individuals with gait impairment and amputees prefer not to stand out from the crowd. The second, excessive asymmetrical gait leads to compensations and earlier degenerative disease such as joint osteoarthritis on the contralateral side due to excessive forces and potential lower back pain (Mena et al. 1981; Winter and Sienko 1988). The third, excessive asymmetry could be associated with increased energy expenditure

(Mena et al. 1981; Engsberg et al. 1991; Lemaire and Fisher 1994). The excessive level of asymmetry considered as a gait disorder (Sadeghi et al. 2000). The level of asymmetry increases with the increased level of amputation (Raggi et al. 2009; Highsmith et al. 2010). The fundamental aim in the development of lower-limb prosthetic devices is to provide symmetrical and a biomechanically efficient gait pattern. Subsequent researches have reported that heavier prosthetic devices assist in the propulsion of the trunk forward (Gitter et al. 1997; Lehmann et al. 1998) without notably increasing metabolic cost (Gailey et al. 1997). This modification could enhance kinetic and kinematic symmetry (Donn et al. 1989; Mattes et al. 2000) for ground-level locomotion in TTs. Hence, the inertial properties of the prosthetic foot directly affect the spatio-temporal symmetry of gait. Interestingly, the research of De Asha (2014) suggests that reduction of the braking effect from the prosthetic foot in the first part of the stance phase could be a more valuable function of the ankle-foot device than late stance energy return (push-off) (De Asha et al. 2014). A hydraulically articulated ankle-foot device increases walking velocity (De Asha et al. 2014) and reduces spatio-temporal symmetry in overground gait (Nolan et al. 2003), so the locomotion will be less symmetrical, but more efficient. More efficient, but asymmetrical gait has an unknown effect on the whole biomechanical system. Considerably more work will need to be undertaken to determine the complete effects on the whole biomechanical system.

Therefore, the examination of gait symmetry should always be methodological and include different biomechanical variables and an examination of the correlation between these variables. The studies of the paragraph above have provided researches for overground gait, but little is known about the effects of prosthetic devices with different mass and functionality on inclined surfaces. Nevertheless, the rehabilitation and prosthetic limb design should lead to the enhanced symmetry between limbs with an improvement of walking speed. The excessive gait asymmetry could affect the biomechanical system. Enhanced symmetry between limbs typically would lead to more efficient (reduced energy expenditure) and safe locomotion.

2.7.2 Kinematics

Kinematics describes rigid body segment motion in space without forces that influence these motions. The observational method of gait analysis was often used by clinicians to examine gait kinematic parameters, but the simplicity of this method can take priority over accuracy (Wall and Scarbrough 1997). However, the accuracy of kinematic parameters improved within the development of motion capture systems and biomechanical modelling. Kinematics of lower-limb joints is widely used in gait analysis studies.

It is critical to emphasise pelvic range-of-motion (RoM) in lower-limb amputees. In TTs compared to able-bodied individuals pelvic a RoM in the frontal plane is amplified with speed (slow to self-selected walking speed) so TTs have to compensate by lifting the pelvis during a swing (Su et al. 2007). This pelvic motion has been described as hip hiking (Michaud et al. 2000) in TTs through the deficiency (dynamic response prosthetic feet have a fore-foot keel which could partly supply dorsi-flexion due to spring properties) of ability to have dorsi-flexion motion on a prosthetic side. This motion has positive and negative outcomes. The positive is that hip hiking increases vertical toe clearance (VTC) during a swing on the prosthetic side (Su et al. 2007) and as an outcome, it is a safe gait pattern. The negative outcome of hip hiking is a rise in metabolic cost and an increase of asymmetry in gait pattern which potentially could lead to contralateral hip osteoarthritis (Kulkarni et al. 1998; Norvell et al. 2005) and lower back pain. However, the research of McNealy and Gard (2008) suggested that use of different prosthetic feet for bilateral trans-femoral amputees (TF) do not have an effect on RoM in the sagittal plane (McNealy and Gard 2008). Postema et al. (1997) also supported that examination of TTs gait had no preference between different prosthetic feet (energy storing feet and conventional feet) (Postema et al. 1997a). To date, researches have not used a hydraulic articulated ankle so prosthetic 'ankle' articulation could have a different effect on the hip motion in the frontal and the sagittal planes. Further studies, which take these variables into account, will need to be undertaken.

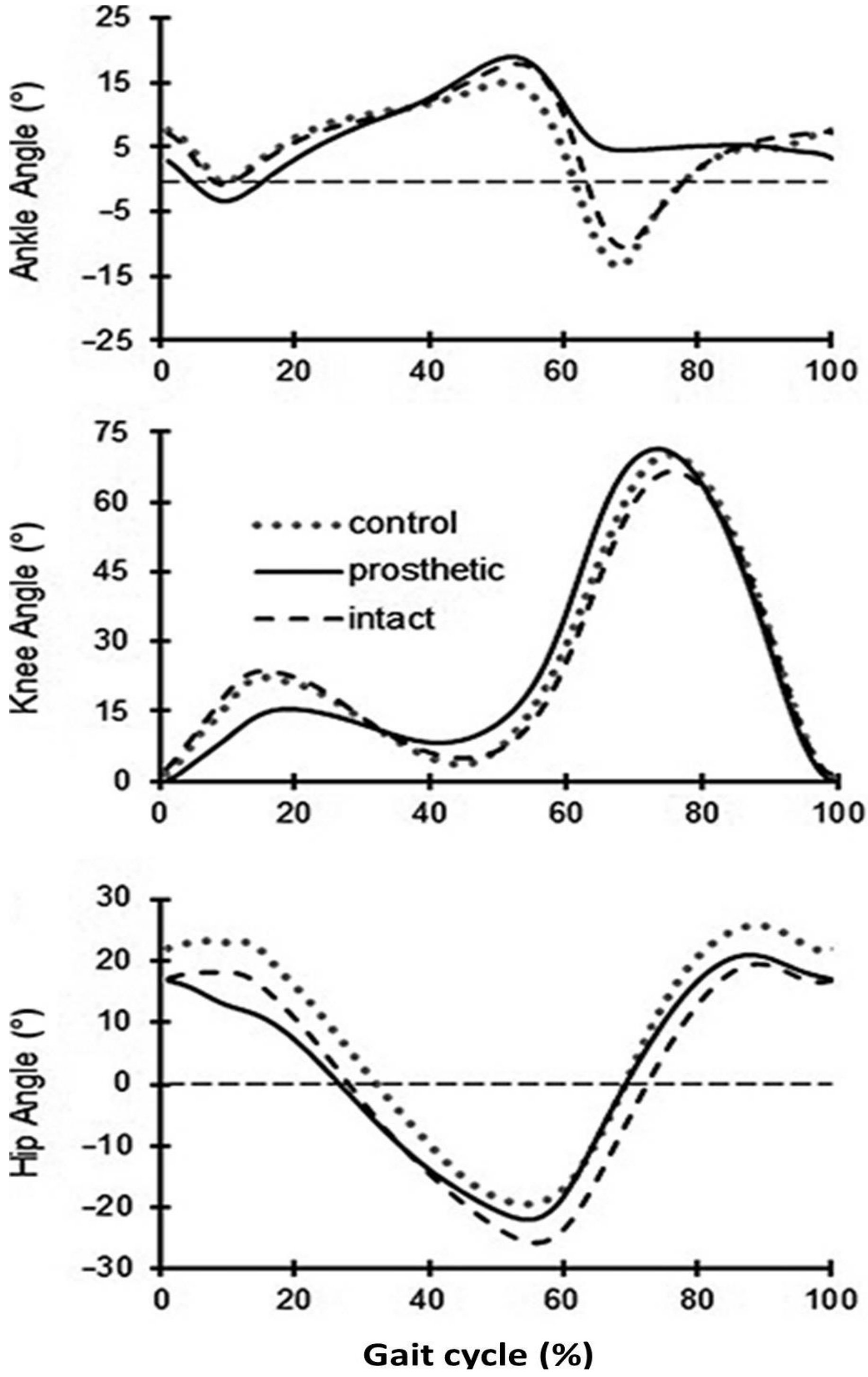

Figure 8. Able-bodied, prosthetic and intact (ankle, knee and hip joint) of TTs flexion-extension (sagittal plane) throughout the overground gait cycle, where is flexion positive angles. Curves are group averages for each limb and normalised to percentage gait cycle. Adapted from Schnall et al. (2014) (Schnall et al. 2014).

Assessment of kinematic differences in lower-extremity amputees have presented a higher risk of falling or fear of falling than able-bodied individuals (Kulkarni et al. 1996; Miller et al. 2001b). The research of Vanicek and colleagues (2009) had a deeper examination of TTs gait by comparing fallers versus non-fallers but did not distinguish stance limb joint mobility between them. The investigation of prosthetic swing limb kinematics indicated that non-fallers have shown increased (residual) knee flexion during the swing phase for approximately 7 degrees with less variability of this period. However, amputee fallers have shown an increased load on the prosthetic side. Interestingly, Vanicek and colleagues compared TTs characteristics to aged adults with muscle weakness and postural instability (Isakov et al. 1992; Vanicek et al. 2009). The difference between groups contrary to the predicted by Vanicek et al. (2009) in the joint moments or powers of the lower-limb system. Possibly, it was due to various types of prosthetic feet (Vari-flex, Multiflex, Ceterus). The various types of prosthetic feet have different weights from Multiflex (375g) (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK), Variflex (700g) and Ceterus (896g) (Ossur, Reykjavik, Iceland) which could have an effect on performance and proprioception of amputees. Further work is required to establish the effect of prosthetics functionality as it might affect the risk of falling. The use of different weight and functionality prosthetic feet is likely to lead to distinct compensations and possibly could have an influence on falls.

The crucial point in TTs is a RoM at the knee joint as a compensatory mechanism. At initial contact, the function of knee flexion (sagittal plane) is to absorb shock in order to reduce the impact on weight-bearing joints (Isakov et al. 1996a) and residuum. The knee extensors work was reduced due to the inertial properties of prosthetic devices. The RoM at the knee joint is between 15˚-18˚ in able-bodied individuals and intact side versus 9–12° in TTs (Powers et al. 1998; Su et al. 2007). Other researches have also supported that residual knee flexion during loading response has reduced compared to the knee of intact side (Winter and Sienko 1988; Sanderson and Martin 1997). Reduction of residual knee flexion during loading response could be related to various factors. Prosthetic socket interface restriction (Isakov et al. 2000). Partly amputated muscle weakness at the residual knee (Winter and Sienko 1988) during eccentric muscle contractions and keeping knee extended to prevent feelings of the knee giving way. Rehabilitation process can also reduce residual knee flexion during loading response. Additionally, contraction of residual limb tissue after surgery could lead to a reduction in RoM due to pain during wearing of the prosthetic socket (Gailey et al. 2008). Also, note that reduction of residual knee flexion during loading response in TT could be the result of muscular co-contractions around the knee in order to stabilise the joint (Segal et al. 2012). Reduction of residual knee flexion leads to an increase of RoM on the intact side (Vanicek et al. 2009), because amputees desire to maintain speed and/or step length.

The able-bodied foot is commonly modelled as a rigid segment. The examination of ankle RoM in able-bodied individuals is presented in a number of researches (Roaas and Andersson 1982; Blanke and Hageman 1989; Kerrigan et al. 1998). For example, Kerrigan and colleagues examined RoM at the ankle in elderly able-bodied individuals in comparison to young adults (Kerrigan et al. 1998). The elderly population has presented reduced RoM at the ankle (Murray et al. 1969; Kaneko et al. 1991). Hence, based on modelling the use of ankle angle can be a valid parameter for gait analysis. On the other hand, the examination of RoM in prosthetic 'ankle' has to be done with care. RoM of the prosthetic 'ankle' is dependent on the type of prosthetic foot design used. In prosthetic ankle-foot, designs the RoM depends on articulation in the 'ankle' and deformation properties of the heel and fore-foot keel during a gait. RoM in 'ankle' is commonly provided by manufacturers. The peak 'plantar-flexion' occurs on the end of the first rocker to ensure the prosthetic foot has maximal contact as it offers better stability (Powers et al. 1994). Prosthetic 'ankle' articulated towards 'plantar-flexion' and heel keel deforms to attain foot-flat. The peak 'dorsi-flexion' occurs at the end of the second rocker, where the prosthetic ankle-foot device optimally should respond according to the gait pattern of an individual. The fore-foot keel of the prosthetic device has to transfer to the third rocker where the final phase 'push-off' occurs. The research of Powers et al. (1994) indicated that improved RoM of prosthetic foot devices advocated RoM reduction on the ankle of the intact side (Powers et al. 1994). However, the research examined a different type of prosthetic feet where the direct comparison could be deceptive. Because, to measure RoM in the prosthetic foot is not appropriate as it would depend on the deformable properties of heel and fore-foot keel, gait pattern and weight of the amputee. To support this, the research of Vanicek and colleagues (2009) did not distinguish prosthetic foot with different RoM/functionality between TTs' faller and non-fallers (p=0.53) (Vanicek et al. 2009) so supports the idea that the assessment of RoM in a prosthetic foot can be misleading.

Kinematic parameters are important measures for gait analysis as they describe joints RoM, velocity, and acceleration of the segments. Kinematics present body segments or joint motion without displaying forces that are applied to these joints. However, kinematic parameters are used to calculate joint kinetics. The repercussion of GRFs, joint moments and muscle powers are critical, it because explains the significance of work which caused that motion (Winter 2005), so could help to elucidate that motion. Further investigation would include an overview with a critical literature review of ground reaction forces (GRF), kinetics, Center of Pressure (CoP) for able-bodied individuals and lower-limb amputees.

2.7.3 Ground Reaction Forces

The ground reaction forces (GRF) data report is an accurate description of gait diagnostic for clinical investigations to assess pathology (Winter 1991; Perry et al. 1992). The GRFs force-time curves (Anterior-Posterior GRF (A-P GRF), Medial-Lateral GRF (M-L GRF) and vertical GRF (vGRF)) are characterised by the effect of the whole body motion in three dimensions. The common examination of GRFs includes peaks, peak occurrence, impulses. The asymmetry in GRFs between limbs of assessed variables is greater with increasing degrees of this pathology in patients with unilateral gait impairment. The GRF data of individual assessed gait could be compared to GRFs of the control group (individuals without gait pathology). To measure GRFs The most commonly used electronic force platforms: AMTI (AMTI Inc., Watertown, MA, USA) and Kistler (Kistler Instruments, Hampshire, UK). Figure 9 (below) illustrates GRFs of TTs and able-bodied individuals over a custom dual-belt treadmill (flat level) with different walking speeds (Giest and Chang 2016). The researchers Giest and Chang (2016) recruited TTs that utilised passive-elastic ankle–foot components.

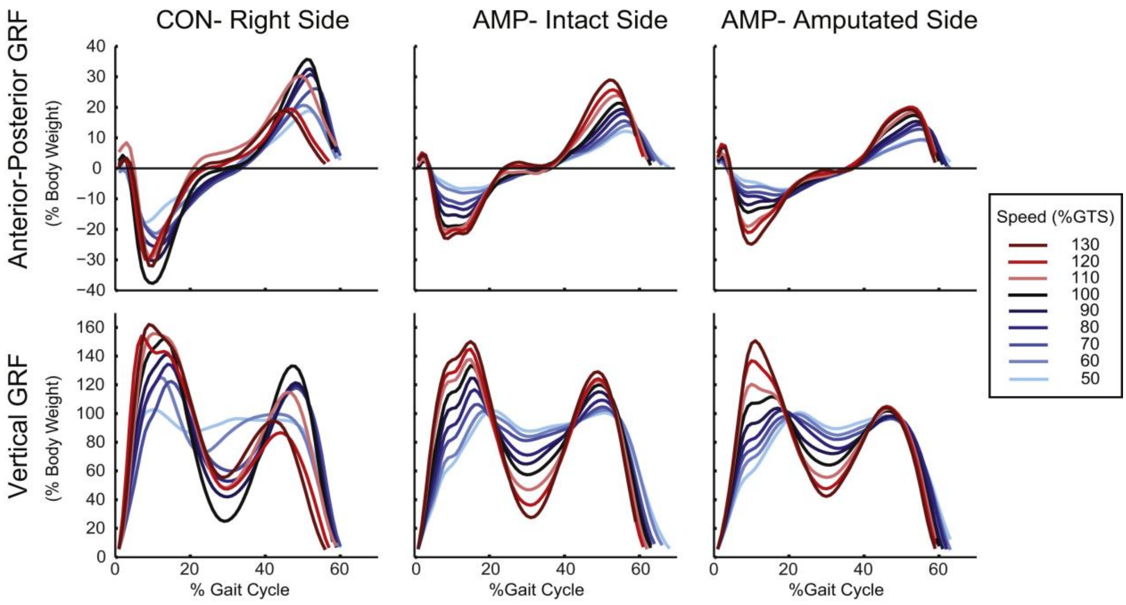

Figure 9. TTs (5 male and 5 female participants) and matched able-bodied individuals (5 male and 5 female participants) ground reaction forces (GRF): anterior-posterior (A-P GRF) and vertical (vGRF) throughout the stance phase. Curves are group averages normalised to percentage gait cycle where the colour of the line illustrates an individual's gait transition speed over a custom dual-belt treadmill (flat level). CON - able-bodied individuals controlled limb, AMP – Intact side – TTs intact side, AMP – amputated side – TTs residual side. Adapted from Giest and Chang (2016) (Giest and Chang 2016).

Prosthetic foot simulates ‘plantar-flexion’ to provide the propulsion of the body forward with swing initiation (Neptune et al. 2001; Neptune et al. 2004). However, the prosthetic foot has reduced push-off compared to the biological ankle and compensated by intact side (Winter and Sienko 1988). Unilateral lower-limb amputees gait has an increased duration of stance phase and load on the intact side (Burke et al. 1978). The study of Gailey and colleagues (2008) presented that GRF on the intact side is 23% greater than the prosthetic side (Gailey et al. 2008), so logic suggests that outcome is a compensatory mechanism. The compensatory mechanism employs the intact side. The compensation mechanism is required to compensate for the missing ankle function. The prolonged stance on the intact side could be due to amputees' difficulty in maintaining balance and stability on the prosthetic side. Although, forces that affect the intact side are dependent on the type of prosthetic device (Agrawal et al. 2015). The prosthetic devices with dynamic response tend to reduce the first peak of GRF on intact side and flexion moment in this limb (Lehmann et al. 1993). The second peak of vGRF increases with speed and becomes more asymmetrical as it is rising faster under the intact than prosthetic side as the prosthesis could not adapt to walking speed changes (Sanderson and Martin 1997). The study of Nolan and colleagues (2003) presented that the increase of contralateral limb involvements with the improvement of walking speed in unilateral transfemoral and TTs is the compensatory mechanism (Nolan et al. 2003). The compensatory mechanism affects contralateral joints: ankle, knee and hip (Nolan et al. 2003). Interestingly, the results of De Asha et al. (2014) study suggest that reduction of the braking GRF on the prosthetic foot could be more beneficial than propulsion (push-off) in TTs (De Asha et al. 2014) which possibly increases asymmetry between limbs. The study of De Asha et al. (2014) presented that the use of hydraulically articulated ankle-foot devices reduces the braking GRF but increases self-selected walking speed in TTs compared to rigid ankle-foot devices. The increase of the self-selected walking speed that is preferred by individuals is due to the most energy efficient locomotion (McNeill A 2002; Figueiredo et al. 2013) and suggests improvement of gait. It can be suggested that walking speed presents quality of gait where moderate level asymmetry between limbs could be accepted. In able-bodied without gait pathology, individuals with vertical GRF asymmetry could be up to 4% (Robinson et al. 1987), so the risk of developing degenerative disease would be reduced relative to the intact side of unilateral lower-limb amputees (Burke et al. 1978; Kulkarni et al. 1998; Hurwitz et al. 2001). Although in the review of Sadeghi it has been shown that some researchers, even recently (Nymark et al. 2005), have evaluated just one limb with the presumption that a second limb would represent equal results (Hannah et al. 1984; Sadeghi 2000). Interestingly, the study of Robinson and colleagues found that anterior-posterior with medio-lateral GRF more asymmetrical than vertical GRF (Robinson et al.

1987). Subsequently, the study of Gailey and colleagues presents if amputees optimally exploit the prosthetic device and have the correct rehabilitation process the risk of osteoarthritis could be diminished (Gailey et al. 2008). Indeed, medio-lateral GRFs have a place but the motion of ankle is limited by the ankle-foot orthosis to sagittal plane motion, and typical ‘ankle’ mechanism of prosthetic devices was limited to the sagittal plane motion. Medio-lateral impulses and forces only influence the propulsion phase within an effect on inversion/eversion of the ankle. Hence, these forces would not be examined further.

As mentioned in the previous paragraph excessive compensations could lead to muscular-skeletal conditions such as back pain which commonly ensues lower-limb amputation, 47.7% are faced with it (Smith et al. 2008). These outcomes have been represented in some studies as an effect of the difference in leg length, reduced hip extensor and back strength, the flexibility of iliopsoas and other amputees’ downsides compared to able-bodied individuals (Gailey et al. 2008; Smith et al. 2008; McGregor and Hukins 2009). Functional characteristics of amputees’ are mostly the consequence of effects in GRF and research of Kulkarni and colleagues found significant ($p<0.05$) difference of GRF between amputees’ with and without back pain (Kulkarni et al. 2005). Nevertheless, this study concluded that the difference in postural muscles effects asymmetry of gait and predisposes TTs to low back pain (Kulkarni et al. 2005). The researchers finalised this study in conclusion that where an outcome is based on the effect of GRF on compensatory mechanisms, logic suggests that the development of secondary conditions as effects of GRF and not only as a result of compensatory mechanisms.

There are many factors that could affect lower-limb amputees GRF. The walking speed is one of the main parameters to assess the quality of gait in individuals (Skinner and Effeney 1985) would have an effect on GRF that would be partly dependant on the prosthesis configurations and alignments and have a critical influence on GRF response in lower-limb amputees gait. Prosthetic components would particularly effect anterior-posterior direction due to the stiffness used for prosthesis keel and heel material (Zmitrewicz et al. 2007). Consequently, the stiffness of the prosthetic heel would affect peak braking force with braking impulse and the stiffness of a fore-foot keel would affect peak propulsive force with propulsive impulse. The articulation in the prosthetic ‘ankle’ would affect braking and propulsion impulses which improve loading symmetry between prosthetic and contralateral limb. However, in the UK amputees’ do not have a prescription of optimal prosthetic components, so the amputees’ rehabilitation process is dependent on the prosthetist’s experience, but not on biomechanical data and detailed components’ properties (Hafner et al. 2002). The optimal lower-limb prosthetic device should correspond to the patient abilities and needs. The selection of prosthetic components (modular system) depends on the patient: age, weight, physical condition, length of the residual limb, and the ratio of stride frequency, stride length, cost and preference. The function of prosthetic devices should aid patients' performance by optimising biomechanics not only in overground gait but also during other daily tasks (inclined surfaces, quiet stance, different walking speeds). Optimal performance of a task would require a prosthetic device that was able to change the functionality accordingly to those tasks, for example, slope ambulation (McIntosh et al. 2006). However, prosthetics such as a microprocessor controlled ankle-foot device that adapts during various tasks commonly prescribed for patients with a high level of activity. Gait analysis could be employed to optimise prosthetic prescription to achieve successful rehabilitation. The examination of energy absorption and release in the prosthetic device during performing various tasks within an effect on a biomechanical system could update prosthetic prescription.

Evaluation and analysis of GRFs are critically important for inverse dynamics calculations (Gordon et al. 2004). Particularly in stance phase when during the first rocker the vertical GRF could be raised up to 1.3 times of body weight and vary from gait velocity (Rodgers 1988). The third rocker of stance phase has time at the second peak of vertical forces which should not be under estimated, as the power generated for propulsing the body forward are above the body weight of an individual. In lower-limb amputees changes in GRF under prosthetic device would have to present on the intact side. The articulation in prosthetic ‘ankle’ would have an impact in anterior and posterior direction variables as peak braking/propulsion, braking/propulsion impulses and braking to propulsion transition point.

2.7.4 Kinetics

Kinetic data reports moment and powers which is an accurate description and examination of gait. To calculate kinetic parameters, an inverse dynamics approach was employed. Kinetic parameters used for clinical investigations to assess gait pathology of lower-limb joints (Winter 1991; Perry et al. 1992). Joint kinetics is the study of the cause of motion that relates to the motion of body segments with associated forces. The force plate works in conjunction with the motion capture system. The interior loads at lower-limb joints are calculated from external GRF data motion data with an anatomically relevant (AR) biomechanical model applied. The AR approach was used for participants’ lower-limb joints. However, to calculate the prosthetic side would be problematic and misleading (Geil et al. 2000; Sagawa et al. 2011). The evaluation of a prosthetic ‘ankle’ joint would not be accurate and consistent between prosthetic devices due to the different stiffness of heel- fore-foot- keel deformity. To deliver an approach that is more appropriate for the calculation of kinetic parameters researches have attempted to model prosthetic devices accordingly to its functionality. To calculate prosthetic foot energy an approach was used (it did not evaluate the prosthetic ankle joint centre) that was introduced by Prince et al. (Prince et al. 1994) moreover, it was later validated and modelled as a Unified deformable segment (*UDS*) by Takahashi et al. (Takahashi et al. 2012). The assessment of lower-limb joint moments and powers compared to control limb joints shows that an increase of difference in assessed variables increases levels of its pathology. Indeed, in patients with unilateral joint impairments, the joint kinetics also can be assessed on the asymmetry between limbs. Figure 10 (AB) illustrates an example of able-bodied individuals (controlled) with standard deviation. Figure 10 (CD) illustrates TTs joint moments and powers group average with standard deviation (Winter and Sienko 1988).

Figure 10 illustrates the gait cycle from initial contact (IC) to IC of the same foot. There 60% presented stance phase which was started from lateral foot IC till toe off (TO) of the same foot. The ankle moment curve (Figure 10 A) after IC shows dorsi-flexors (negative) moment, which prevents foot 'slop' (foot-flat attained too quick). The curve notifies that throughout the stance is mainly active planter-flexors muscles (triceps surae) which are increasing activity at the end of the stance 'push-off'. The increase of plantar-flexor moment during loading response is linked to increasing plantar-flexion at IC. After 'push-off' a dorsi-flexor moment occurs in order to lift the toe from the ground and provide toe-ground clearance. Reduction of the dorsi-flexor moment is linked to the reduction of the first rocker. The knee moment curve has the first peak extensor (positive) which acts in order to prevent limb collapse. At the end of the stance, before toe-off, the knee moment curve commonly shows the involvement of the flexor muscles, so the limb is pulling though remains of the stance phase. In the following swing phase, the knee moment curve shows at the begin extension (limit knee flexion as swing from the hip) which followed by flexion (before knee reaches full extension). The examination of joint kinematic during swing phase in locomotion presented a non-significant muscle activity with passive power flow distally through the joint (Siegel et al. 2004). The hip moment curve the early stance involve extensor muscles then flexor muscles to reduce limb velocity before IC. The hip moment is a critical parameter in gait assessment as controls lower-limb and balance of the trunk (Gordon et al. 2004; Chapman 2008; Winter 2009).

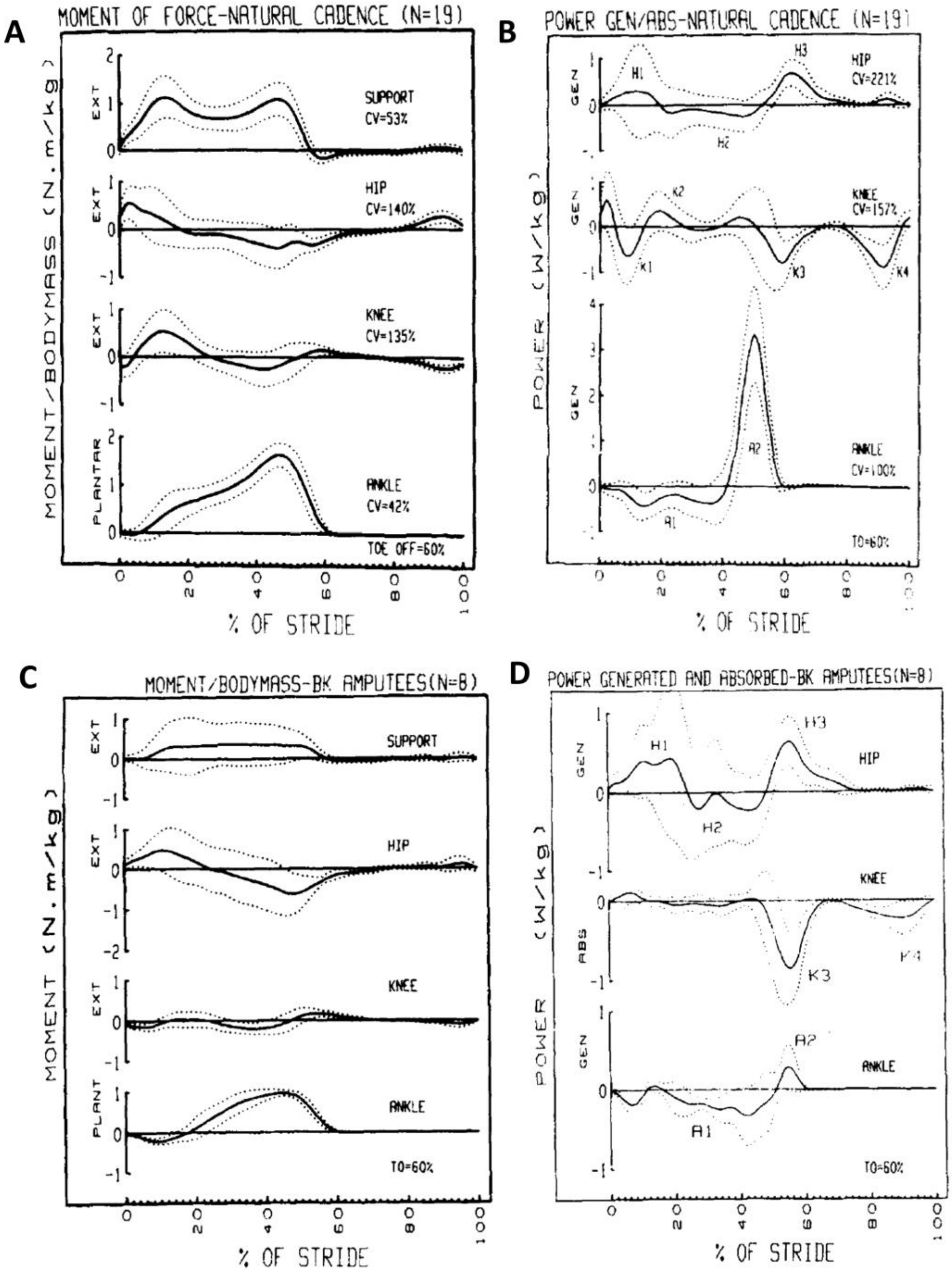

Figure 10. A and B curves - represents sagittal plane mean data of able-bodied individuals or controlled (19 participants): support moment, hip, knee and ankle joint moments and powers C and D - curves represents sagittal plane mean data of trans-tibial amputees prosthetic limb (8 participants): support moment, hip, knee and ankle joint moments and powers. Curves are group averages normalised to percentage gait cycle Adapted from Winter and Sienko (1988) (Winter and Sienko 1988).

Muscle contracting has used the term ‘mechanical power’. The calculation of joint power is performed, the external joint moment multiplied by the angular velocity of the joint. Moment power was a product of the proximal joint moment was to provide the net moment of the force times angular velocity of the assessed joint.

$$P=M*\omega_s \qquad \text{Equation 3}$$

M - a moment in the sagittal plane that applied at the proximal end (N*m). ω_s - angular velocity at the assessed joint (rad.s^{-1}) where is a displacement of one segment in relation to another segment over a period in the sagittal plane.

$$W=\int_{t1}^{t2} P dt \qquad \text{Equation 4}$$

Negative and positive joint work (power integrals) were examined as independent variables to specify eccentric and concentric work when lower-limb joint absorbs or return energy during the period (t_1 and t_2). t_1 and t_2 are time for integration between distinct periods of times. The power is positive if two segments of the joint are moving in the direction of the concentric muscle contraction. The power is negative if the segments of the joint are rotated away from the direction in which the muscle is pulling, and identified as an eccentric contraction. If the joint is not moving there is an isometric contraction, the result of the angular velocity, and the power will be equal zero. The power integrals describe whatever muscles are being used to perform external work.

In overground gait TTs demonstrate a reduction of residual-knee loading response flexion within joint reduction moments (peak, impulse), peak power and work compared to intact side (Czerniecki et al. 1991; Gitter et al. 1991; Sanderson and Martin 1997; Powers et al. 1998; Sagawa et al. 2011). Increased walking speed reduces temporal asymmetry but increases hip power generation on the intact compared to prosthetic sides with increases of asymmetry (Silverman et al. 2008). The increased intact side involvement is likely to be the result of compensatory-protection mechanism as an increase in walking speed should increase joint contributions (De Asha et al. 2013b). To compensate for the lack of prosthetic-side ankle power generation, the hip flexor involves a pull off strategy (McGibbon 2003). However, during slope descent, the requirement of propulsion ('push-off') is reduced due to increased potential gravitational energy. Slope descent involves an increase of walking speed control rather than propulsion, so the effect of the speed on TTs is unknown and requires investigation.

2.7.5 Centre-of-Pressure

The Centre-of-Pressure (CoP) progression in normal able-bodied gait is throughout the stance phase from the lateral border of the heel at the initial contact to hallux or big toe at toe-off. During locomotion, the CoP progress as the shank rotates over the support foot with the transfer of the CoM forward. The CoP progression defines as the origin of the application of the ground reaction force vector (Winter 2009) and reflects control of the whole body CoM forward motion (Kirtley 2006a). The CoP progression was presented as a measurement of neuromuscular control within the posture and gait of an individual (McPoil et al. 1989; Chesnin et al. 2000). The CoP progression beneath the foot is used to recognise how an individual controls balance, and what is the functionality of the contacted (with the ground) foot. Hence, the functionality of the prosthetic device used and/or the effectiveness of treatment received can be indicated by the CoP progression. The examination of the CoP velocity progression could identify gait efficiency. A notable difference in the CoP progression between lower-limb amputees and able-bodied individuals have been presented in numerous investigations (Jones et al. 2005; Schmid et al. 2005; Kendell et al. 2010).

The time was kept longer when CoP remained under the mid-foot region of the prosthetic foot compared to the intact or controlled limb (able-bodied) (Breakey 1976; Engsberg et al. 1993; Schmid et al. 2005). The research of Schmid et al. (2005) presented that the CoP velocity beneath the heel and mid-foot regions of the prosthetic foot was kept longer than on intact side but kept shorter beneath the fore-foot region relative to the percentage of stance (Schmid et al. 2005). These suggest that the CoP progression prolonged stance of the intact foot compared to the controlled limb and indicated the involvement of a compensatory mechanism. However, the research did not find the difference between controlled and intact limbs. Thus, slower CoP velocity beneath the heel and mid-foot region of the prosthetic limb is consistent with unilateral amputees' feedback as 'climbing over' or 'dead' spot a during the period of early or mid stance phase. There CoP forward progression was disrupted (i.e. reduced aggregate negative CoP displacement) beneath the heel and mid-foot regions when transferring the CoM of the whole body over support prosthetic foot (Schmid et al. 2005; Winter 2009; De Asha et al. 2013a). Use of hydraulically articulated ankle-foot device have presented the delay but not eliminate the 'dead spot' (De Asha et al. 2013a). The CoP progression is controlled and depends on the functionality of the lower-limb device (Hafner et al. 2002) throughout three rockers (from initial contact to toe-off). There prosthetic device components and prosthetic alignment have an influence, predominantly during the first and second rockers (Schmid et al. 2005; De Asha et al. 2013a). From the points mentioned above the critical difference between intact and prosthetic lower-limb amputees and able-bodied individuals can be established. Compensations in unilateral amputees could vary according to prosthetic foot designs. Based on factors mentioned above the CoP progression could be used as a reference for assessment of amputees gait.

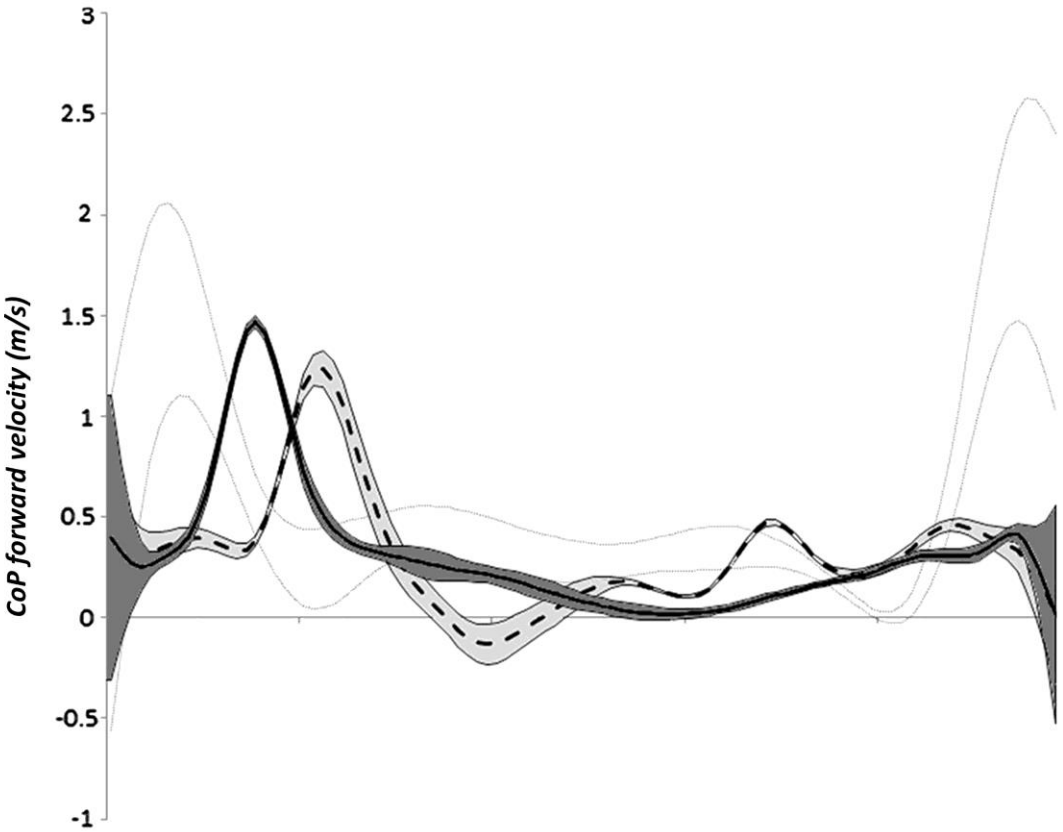

Figure 11. Mean ±SD (10 trials) of CoP forward velocity (normalised to stance phase), an *Echelon* (hydraulically articulated ankle-foot attachment) - (solid line/dark shading); *Esprit* (rigid ankle-foot attachment) - (broken line/light shading). Mean ±SD of CoP forward velocity in able-bodied individuals (dotted lines). Adapted from De Asha et al. (2013) (De Asha et al. 2013a).

The stance remained longer on the intact side not as result of pain (no problems with the residual limbs) or prosthetic device (accustomed prosthetic) (Hurley et al. 1990; Torburn et al. 1990; Schmid et al. 2005), but likely due to the ability of amputees to maintain balance better on the intact side. The transition (initial double support) from intact to the prosthetic side is longer than from prosthetic to the intact limb (Schmid et al. 2005). Hence, the amputee's slower transfer of body weight on the prosthetic side is likely to be due to pertrubated sensomotory feedback. So TTs have a longer braking period than propulsion (Seliktar and Mizrahi 1986). Prosthetic foot design plays a critical part in CoP progression. Figure 11 illustrated that negatively directed CoP velocity during mid-stance was greater for rigid compared to hydraulically articulated ankle-foot attachment (De Asha et al. 2013a). The research suggested that CoP velocity during mid-stance effected by weight acceptance when the heel keel deforms to ‘planter-flex’ under load with transfer onto the fore-foot keel as the CoM moving forwards over the foot (Schmid et al. 2005; De Asha et al. 2013a). De Asha et al. (2013) findings, indicating the use of the hydraulically articulated ankle-foot attachment attenuated negatively directed CoP velocity during mid-stance, as result of ‘ankle’ functioning (De Asha et al. 2013a). Disrupted CoP progression could lead to complications fo control dynamic balance effectively. In the research of Torburn et al. (Torburn et al. 1990) the CoP progression over a prosthetic foot was faster during single-limb-support which could be a result of the prosthetic foot design. The research of De Asha et al. (De Asha et al. 2013a) indicated that variability over ten trials of the CoP progression presents intra-subject consistency and dependents from prosthetic foot design (Figure 11). There CoP forward velocity of able-bodied individuals was faster at the heel and toe regions and almost consistent with low variability during the single-limb-support phase. The fluctuation during single-limb-support was lower for hydraulically articulated ankle-foot device compared to the habitual ankle-foot device due to hydraulic consistent articulation rate and not effected by contralateral limb. As a result, the hydraulically articulated ankle-foot device provides a mode of uniformed transition over the single-limb-support phase. The first peak of the CoP forward velocity was fastest for able-bodied compare amputees with the hydraulically articulated ankle-foot device or habitual ankle-foot device. This delay was likely a result of the heel’s carbon fiber deformation which allows the prosthetic foot to simulate ‘plantar-flexion’. The last peak of the CoP forward velocity was shown that able-bodied ankle provides ‘push-off’ but prosthetic devices could not generate increase CoP forward velocity (Figure 11).

A widely used theory was proposed by Hansen and colleagues (Hansen et al. 2000). The CoP curve was transferred from the laboratory-based coordinated system to a local coordinate system of a shank and has a ‘roll-over’ shape. Where different lower-limb prosthetic devices had to eliminate the model segments and joints. To support this theory it was proposed, that different types of prosthetic devices require different alignments (Hansen et al. 2000). This theory was used by a number of researchers (Curtze et al. 2009; Ren et al. 2010; Gruben and Boehm 2014). The limitation of this theory is the ‘roll-over’ shape has to fit a curve to disrupt the limited number of CoP in some parts of its progression which may have to be to distinct from each other. Anecdotally, to justify that limitation it is possible to fit a ‘roll-over’ ‘best fitted’ curve on a square or triangle due to the low sampling rate so that it changes the implication of CoP use and various outcomes with the use of footwear on the prosthetic side. Additionally, Curtze and colleagues assessed the effect of roll-over shape of the prosthesis on CoP of the intact side which did not show any sign of roll-over shape, however CoP curves were individual (Curtze et al. 2011). Moreover, the CoP disruptions in the second rocker could have a direct connection with the ‘dead spot’ or ‘climbing over’ so the application of this theory has to be used with care.

The muscles and ligaments of the foot provide a natural sequence during locomotion with the underlying effort of maintaining postural stability and balance during the stance phase. The importance of the ankle during stance phase in locomotion was emphasised, that the reduction of ankle strength and/or range-of-motion (RoM) increases the risk of postural instability. Postural stability in lower-limb amputees is dependent on prosthetic design (Vrieling et al. 2008) and limited proprioception from the prosthetic foot (Vickers et al. 2008). The use of ankle-foot design of the prosthesis has improved stability and maintaining static/dynamic balance (Buckley et al. 2002). Postural stability and static/dynamic balance have a significant impact on energy consumption as one of the main considerations for efficient locomotion. The ankle-foot functionality throughout the stance phase provides safe and secure weight bearing of the whole bodyweight (Bateni and Olney 2002). The articulation of ankle-foot during stance phase (from foot contact till toe off) could be described as ‘roll-over’ that affects the smoothness of sinusoidal CoM progression. Disruption in ‘roll-over’ progression along the plantar surface of the foot during the stance phase could effect the smoothness of the CoM progression. In amputees during overground gait, the CoP forward progression being delayed (disrupted) which suggests that prosthetic ‘ankle’ insufficient mobility with articulation rate and/or inadequate heel- fore-foot- keel deformation and recoil properties.

This section explains and discussed the findings of published papers which investigated able-bodied and amputee overground gait. The number of studies has been limited by the separation of variables associated with locomotion. Further research should expand the previous work to address the feasibility of correlation between spatio-temporal, kinematic, kinetic and CoP parameters to investigate the biomechanics of gait correctly. Moreover, the CoP parameter is very sensitive as foot scuff during the first rocker could have a place. This part of CoP progression has to be investigated with care due to the high possibility of error. The prosthetic foot design has a direct effect not only on the CoP progression, but also could affect examination of the inverted pendulum motion which would aid understanding of body transition during single-limb-support. The examination between prosthetic foot designs in the majority of published papers was performed to assess RoM between prosthetic and able-bodied feet. The prosthetic foot in these studies is typically modelled as an intact or able-bodied limb. These examinations presented a direct comparison between prosthetic, intact and able-bodied (control) limbs. Despite this, future assessment should consider different prosthetic foot models as prosthetic feet do not have the same functionality as human. Section 2.8 below describes and discusses the biomechanical measures of lower-limb amputees and able-bodied controls with relevance to use during slope descent.

2.8 Slope ambulation of able-bodied individuals versus unilateral trans-tibial amputees

In everyday life it is important to be able to adapt gait to environmental changes, stairs, and slopes. Lower-extremity amputees often face the task of slope ambulation in their daily activities (McIntosh et al. 2006). This requires alterations in the gait pattern compared to overground gait (Smith et al. 1998). This concept has recently been challenged by Sheehan and Gottschall's studies demonstrating that slope ambulation has a higher risk of falls than stairs with the same angle of inclination (Sheehan and Gottschall 2012). Slope ascent or descent involves adaptation of the gait pattern according to the mode of ambulation (Leroux et al. 2002). If slope ascent requires more effort than overground gait, it might be expected that slope descent would require reduced effort, but this is incorrect as the human biomechanical system cannot conserve energy without dissipation. The experimental data of early research confirms that slope descent is more demanding than slope ascent (Macfarlane et al. 1991; Sin et al. 2001). Downslope gait has an effect on balance for able-bodied healthy individuals during the stance phase and affects lower-limb joints range-of-motion with shorter step length and faster walking velocity (Sun et al. 1996). The ankle mechanism plays a critical role in the achievement of human's natural bipedal locomotion. TTs have to modify gait due to the absence of an ankle function, so it has to be compensated by other joints (Winter 1980). This is not only due to the absence of muscles that provide a natural sequence in locomotion but also with the underlying effort of maintaining postural stability and balance during locomotion. During slope descent, the compensatory mechanism delivers adaptation in gait pattern according to the prosthetic ankle-foot device. Slope descent compared to overground gait potentially decreases dynamic balance (anterior-posterior and medio-lateral direction) compared to overground gait (Gottschall et al. 2011). To achieve safe and efficient gait pattern requires different functionality from the prosthetic ankle-foot device. The instability in TTs stance phase is partly dependant on prosthetic design (Vrieling et al. 2008) and limited proprioception from the prosthetic side (Vickers et al. 2008). The importance of prosthetic ankle-foot device functionality rises on a slope due to the potential decrease of dynamic balance (anterior-posterior and medio-lateral direction) compared to overground gait. In amputees, postural stability and a reduction of balance equilibrium correlate to proprioception from the amputated side. To enhance stability requires attaining foot-flat quicker and the prolonged base of support. To emphasise the importance of the ankle in locomotion, some studies have shown that the reduction of ankle strength and RoM increases the risk of postural instability (Bennell and Goldie 1994; Bok et al. 2013). Lower-limb amputees' have poorer static and dynamic balance control than able-bodied individuals (Buckley et al. 2002) which is related to reduced proprioception from the amputated side. This limited proprioception from the amputated side causes an increase in falls for amputees' compared to able-bodied individuals (Kulkarni et al. 1996; Miller et al. 2001b; Vanicek et al. 2009). However, maintaining balance during the stance phase is one of the main objectives for lower-limb amputees'. An examination of TTs gait indicates a reduction in speed and cadence with less support time on the prosthetic compared to intact side (Vickers et al. 2008). The increase of support time on the intact limb may lead to developing secondary physical conditions such as degenerative joint disease (e.g. osteoarthritis) and/or lower back pain (Kulkarni et al. 1998; Gailey et al. 2008). Shorter swing phase of the prosthetic limb possibly due to inertial properties of the prosthesis or may be due to a simple result of the amputee's increased the stance time on the intact limb (Breakey 1976; Isakov et al. 2000). However, studies of overground gait presented that manipulation of the inertial properties of the prosthesis affected the step length, walking velocity and symmetry of gait (Mattes et al. 2000), where excessive asymmetry may also lead to secondary physical conditions (Gailey et al. 2008) and an increase of energy costs (Selles et al. 2004). Thus, physical conditions affected TTs as a result of the frequent involvement of compensatory mechanisms to fulfil the role of the amputated ankle.

The treatment regime of lower-limb amputees involves gait training. Once patients are comfortable walking on ground level surfaces, gait training continues to stairs, curbs, and ramps, and uneven terrain (Pohjolainen et al. 1990; Sapp and Little 1995). Training for ascending and descending slopes are motor tasks that have been recommended in rehabilitation (Vrieling et al. 2008). Specific gait training is important to ensure that patients achieve the appropriate biomechanics of gait. Parallel bars for gait training on ramp ascent or descent are not necessarily used by physiotherapists, the patient can start by having assistance from a therapist. Ramp descend is problematic for amputees with conventional ankle-foot devices due to deficiency of 'plantar-flexion'. Physiotherapists recommend that during ramp descent training lead with the residual limb (Gailey and Clark 1992). The prosthetic foot during ramp descent seek foot-flat followed by pylon forward rotation, so the knee would flex, and BW would fall posteriorly to the knee. To reduce knee flexion amputee's are advised to reduce step length. Ramp descent training also involves prevention of catching or tripping on the toe during swing phase so the patient should ensure residual knee flexion during the swing phase. The patient repeats these tasks under the supervision of a physiotherapist to ensure a correct gait pattern. Ramp ambulation could include further challenges such as stops, starts, change in velocity, or step length. Clinicians then make the decision for discharge based on the criteria of independence for ambulation on inclined surfaces (Highsmith et al. 2014; Highsmith et al. 2016).

A well-known method used to assess the quality of gait was ‘freely chosen’ as self-selected walking speed. Lower-limb amputees walk slower compared to able-bodied individuals (Boonstra et al. 1993). The restriction of ankle motion would reduce the gait speed on a level ground surface (Murray et al. 1984; Kirtley et al. 1985). The function of the ankle contributes to an optimal gait pattern to smooth sinusoidal transfer of CoM with minimal metabolic energy cost (McNeill A 2002). Safe slope descent compared to overground gait requires more control with greater peak braking but smaller propulsion GRFs (Kuster et al. 1995; Redfern and DiPasquale 1997; Lay et al. 2006; McIntosh et al. 2006) as result of increased gravitational energy (Chapman 2008). There ankle negative work in early stance increases to provide controlled, safe slope descent (Lay et al. 2007). Thus, the function of the prosthetic foot is a primary consideration during stance phases as it provides secure weight bearing of the whole body mass (Bateni and Olney 2002). This importance increases during slope descent in early and mid stance in order to control body transition. The concern of lower-limb amputees throughout the rehabilitation process to reduce negative after-effects of stance and swing phases which could lead to a secondary physical condition. The investigation of prosthetic device functionality is critical for lower-limb amputees to prevent falls and secondary physical conditions with the further benefit of reducing healthcare costs and improving the quality of life for amputees’.

To achieve safe slope descent requires control of kinetic energy, so the muscles of the lower-limb system have to contract (stretch) with a limited return of energy. Kinetic energy control could create a higher risk of injury due to short muscle lengthening (Chapman 2008). To reduce the muscle contraction during slope descent, the step length is reduced (Kawamura et al. 1991; Sun et al. 1996). Lower-limb amputees reduce the step length to reduce hip extension when ascending or descending slopes which is a result of limited proprioception and reduced contribution of force on the prosthetic side (Vrieling et al. 2008). The findings of Vrieling and colleagues confirms that slope ambulation is a challenging task for TTs as conventional lower-limb prosthetic devices are set for a level ground surface (Vrieling et al. 2008). Hence, amputees would have to remodel their gait pattern to correspond to the environment, and prosthesis functionality as prosthetic foot device is not capable of adapting to inclined surfaces as the devices used were designed for overground gait with self-selected walking speed. There, TTs' adapt in correspondence with limited knee flexion as a result of partly amputated posterior flexion muscle (gastrocnemius). Contradictory, Pandian and Kowalske alleged that TTs could ambulate ramps without difficulties. However, side step or diagonal walking was recommended (Pandian and Kowalske 1999).

TTs may experience difficulties in slope walking as a result of amputation of the ankle joint, muscles, and nerves. Prosthetic foot devices have different properties than a biological foot. Reduced 'ankle' range of motion in prosthetic devices compared to able-bodied during stance phase in TTs leads to compensations (Winter and Sienko 1988). During ramp descent, TTs normally have both (residual and intact) limbs straighter when landing on the ground with more extended hip and knee angles (Fradet et al. 2010). At the initial contact on the intact limb, the hip and knee are extended to provide a longer effective limb, because the trail prosthetic foot can not 'dorsi-flex' at the late stance compared to able-bodied (controlled) ankle. There the prosthetic foot 'dorsi-flexion' restricts the body from lowering and causes premature knee flexion resulting in an early heel rise (Torburn et al. 1990). The extended position of the residual limb would force the intact limb to land on the ground from a higher position which increases the knee flexion of the intact limb at loading response. The intact limb ankle compensated to prosthetic (rigid and single axis) ‘ankle’ feet at initial contact by reduced dorsi-flexion and by increased plantar-flexion at toe-off compared to the able-bodied individuals (Vickers et al. 2008). At initial contact on the residual side, the hip and knee more extended compared to able-bodied as result of shorter step on the prosthetic side which makes easier to transfer BW onto prosthesis to lower the body down the slope (Vrieling et al. 2008; Fradet et al. 2010). Their shorter step length reduces the height difference that requires adaptation to control gravitational potential energy (Chapman 2008). To attain foot-flat, the prosthetic foot has to ‘planter-flex’ (deformation of the prosthetic ‘heel’) to provide a stable position for the BW transfer, where there is a delay of attainment of foot-flat would lead to compensation by an increase of loading response knee flexion (Vickers et al. 2008). The loading response knee flexion increases compared to able-bodied as result of shank/pylon pulling forward to establish foot-flat quicker if the prosthetic ‘ankle’ articulation does not allow it.

Slope descent has the increased risk of slips (Redfern and DiPasquale 1997). The risk of slipping correlates to anterior-posterior (McFadyen and Winter 1988) and medio-lateral (Gottschall et al. 2011) stability and is higher on uneven ground compared to overground gait (Redfern and DiPasquale 1997; Sheehan and Gottschall 2012). Some studies during the examination of gait on inclined surfaces do not mention friction coefficient (Lay et al. 2006; Fradet et al. 2010; Major et al. 2014). To reduce slipping, the surface of the slope requires a coefficient-of-friction and this coefficient should be increased with the increased incline as the risk of slipping is directly dependant on the level of slope inclination (McVay and Redfern 1994). The peak of shear forces accrues at approximately 19% of stance phase for ramp descent (Cham and Redfern 2002) which is the second rocker so the prosthetic ankle-foot articulation (function) could have an effect. To avoid slipping and an eventual fall the friction force has to be greater than the shear force to provide vertical GRF. Hence, lower-limb amputees could establish secure foot contact with transfer to the second rocker with a higher level of confidence.

Changes in walking speed correlate to anterio-posterior GRF impulses (Peterson et al. 2011). Hence, to control slope descent speed require increase braking forces compared to overground gait. The study carried out by Franz et al. (2012) was presented that the leading limb has an increased braking impulse during ramp descent compared to overground gait (Franz et al. 2012). Several sources have identified the increased peak braking, and reduced peak propulsion associated with ramp descent compare to overground (Lay et al. 2006; McIntosh et al. 2006). The anterior-posterior GRFs are dependent on the ankle-foot functionality (Agrawal et al. 2015). Prosthetic ankle-foot articulation rate with properties of heel/fore-foot keel, in conjunction with a gradient of the approached surface and walking speed would affect the produced anterior-posterior GRFs. Certainly, medio-lateral GRFs have to be mentioned as provide information about balance during the gait (Birrell et al. 2007). The maintenance of medio-lateral balance is a consideration in lower-limb amputees gait because it is activated by muscles (Kuo 1999; Donelan et al. 2004). Medio-lateral GRFs change according to various gait velocities and affected by muscles contribution (>92%) for overground gait (John et al. 2012) which was likely affected during slope descent. Produced medio-lateral GRFs effects of speed during a ramp descent are unknown and especially for various lower-limb prosthetic devices. Analysis of kinetic variables during slope descent is crucial for lower-limb amputees' as delay in applying weight bearing of the body weight, on the prosthetic side could lead to a reduction in stability.

Gait on inclined surfaces leads to increased RoM in the lower-limbs (Lay et al. 2006). Designs of the majority of commercially available lower-limb prosthetic devices have a common drawback of constrained RoM that effects slope ambulation. During slope descent, TTs reduce mid-stance residual knee flexion in order to maintain CoM position for forward progression (Vickers et al. 2008). The TTs of Vickers et al. (2008) research wore rigid 'ankle' (SACH) and single axis foot devices. Thus the knee compensates as result of constrained 'dorsi-flexion' in slope descent. Another research of Vrieling et al. (2008) did not present major change in mobility of the prosthetic ‘ankle’ during slope descent, however pointed out that amputees’ with the flexible ‘ankle’ have a ‘dorsi-flexion’ angle similar to able-bodied individuals during the third rocker (Vrieling et al. 2008). The research of Vrieling et al. (2008) distinguished stiffness between relatively flexible (C-walk (Otto Bock, Germany), Quantum Foot (Hosmer Dorrance, Campbell, California), Multiflex (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK), Griessinger (Otto Bock, Germany)) and rigid (SACH foot, S.A.F.E. II (Foresee Orthopedic)) prosthetic feet.

The research of Fradet et al. (2010) has questioned the benefits of adaptable ankle-foot device (Proprio-Foot (Ossur, Reykjavik, Iceland)) during ramp descent, but pointed out that TTs with such a prosthetic foot felt safer with better support and reduced stress at the residual knee, however, this was not supported by results (Fradet et al. 2010). Interestingly, Fradet and colleagues used an inclination of 7.5˚ but did not find any significance in the results and suggested to increase the level of inclination to 15° according to McIntosh et al. (2010) research (McIntosh et al. 2006). An increase of slope gradient of 7.5˚ would be considerably steeper than the inclination that was used to guide the design for building disabled ramp access (BS 8300:2009+A1:2010). Another source also suggested building disabled ramp access at a maximum slope gradient 1:12 (4.8˚) (Alderson 2010). It can, therefore, be assumed that the lower-limb prosthetic devices have to provide the required ‘ankle’ motion with resistance (rate of articulation) towards ‘plantar-flexion’, or ‘dorsi-flexion’ depends on the approached surface inclination. The study of Klute et al. found that restricted RoM in the ankle of able-bodied individuals may create a similar compensatory mechanism to TTs in overground gait (Klute et al. 2001). The assessment of ramp descent is critical as gait on surfaces with an incline has a higher risk of slipping and falling relative to overground gait (Redfern et al. 2001). The TTs also have a high risk of falls due to musculoskeletal impairment (Miller et al. 2001b), for example, due to partially lost ankle function, such as plantar-flexion, which is known to control balance during locomotion (Neptune and McGowan 2011). Hence, little is known about the effects of the restricted ankle on able-bodied individuals and prosthetic ankle-foot articulation types on TTs during the ramp descent.

Although, the current study does not assess lower-limb amputees' falls it is crucial to understand their prevention for the rehabilitation and post-rehabilitation period. Falls of lower-limb amputees could lead to detrimental outcomes and have been widely investigated (Blake et al. 1988; Kulkarni et al. 1996; Miller et al. 2001a). Prevention of falling in the amputee population is an important on-going focus in the rehabilitation process. Correct rehabilitation process could involve improvement of the proper muscle activation as well as physical ability, which involves vision and physical adeptness (Kulkarni et al. 1996; Esquenazi and DiGiacomo 2001). The rehabilitation process aims to remodel the gait pattern to ensure maximal functional mobility and safety. The rehabilitation involves optimising symmetry during swing and stance phases between both prosthetic and intact limbs (Baker and Hewison 1990; Isakov et al. 1996b; Soares et al. 2009). The primary safety concern in rehabilitation is to ensure optimal swing foot clearance of the prosthetic foot to avoid trips. Another safety concern is the amputees' ability to adapt to the physiological changes, prosthesis, surrounding environment, and experience (Miller et al. 2001b) which inability could lead to slips and trips and eventually fall. The research of Buzcek and colleagues presented that impaired individuals required greater coefficient-of-friction (CoF) compared to able bodied individuals requirements are when rising on slopes to prevent slips (Buczek et al. 1990). Unilateral lower-limb amputees require greater coefficient-of-friction (CoF) due to reduced of applied forces on the prosthetic side compared to the unaffected side (Nolan et al. 2003). Hence, the rehabilitation process has to focus to improve load on the prosthetic limb. On a slope frictional requirements increases and rise with an increase of gradient inclination (Redfern and DiPasquale 1997). During slope descent, shear forces increase at initial contact and increase through a significant part of the stance (McVay and Redfern 1994). In rehabilitation, when slope descent is approached to increase friction between the shoe and the ground foot-flat should be attained quicker. Therefore, amputees were instructed to load the prosthetic limb in order to reduce the shear forces between the shoe and the ground as a posterior shear force achieves a maximum during the loading response. To prevent slips, the shear forces should not exceed the friction between the shoe and the ground (Redfern et al. 2001). Aruin et al., (1997) highlights the intact limb in lower-limb amputees plays a significant role in maintaining postural control in amputees but prosthetic limb was unresponsive (Aruin et al. 1997). Patients faced difficulty when required to modify GRFs to changes in surroundings plus the functionality of the prosthetic device (Nolan et al. 2003; Silverman et al. 2008). The fear of falling and the restraints of the prosthetic device function could also have an adverse effect on the rehabilitation process (Russek 1961). The fear of falling should not be underestimated as it could have an adverse effect on the quality of life, rehabilitation process (Howland et al. 1998) which could prolong their stay in hospital (Bates et al. 1995). Falls could lead to injuries related to residual limb trauma (Behar et al. 1991), fractures (Gonzalez and Mathews 1980; Gooday and Hunter 2004) and others. Consequently, the evaluation of fall-related factor's is an important objective for biomechanical researchers to ensure the safety of amputees during their approach to inclined surfaces.

Trans-tibial amputation creates a musculoskeletal disbalance which creates a compensatory mechanism that has a subsequent effect on the gait pattern (Gamble and Rose 1994) which could lead to a secondary physical condition. The secondary physical condition of lower-limb amputees is primarily due to limited (by prosthesis) functionality from the amputated side. This activates a compensatory mechanism in the lower-limb system. In order to achieve a safe, efficient gait pattern, with minimal energy expenditure, amputees' use a compensatory mechanism. The compensatory mechanism increases asymmetry in the load and stance time between lateral and ipsilateral limbs (Burke et al. 1978). The compensations lead to step length, (Zmitrewicz et al., 2006) trajectories in CoP (Hansen et al. 2004), GRFs presentation (Nolan et al. 2003) or kinetic parameters (Silverman et al. 2008) asymmetry in lower-limb amputees for overground gait. A limited number of researches have assessed the CoP progression during slope descent in lower-limb amputees, so examination of prosthetic ankle-foot types could provide a deeper insight into the development of secondary physical conditions as CoP controls whole-body CoM forward progression (Schmid et al. 2005; Winter 2009). Prosthetic alignment can also have an effect on asymmetry distribution of forces between the lateral and ipsilateral limbs (Gailey et al. 2008). The compensatory mechanism for TTs during slope descent is a distinctive form of overground gait. This is mainly due to an increase of ipsilateral knee flexion that compensates for limited knee flexion ability on the lateral side to provide safe slope ambulation but without differentiating the type of lower-limb prosthetic devices used (Vrieling et al. 2008). The compensatory mechanism during slope descent in TTs with Proprio-Foot (Ossur, Reykjavik, Iceland) has been investigated in the study of Fradet et al. where moments and powers of the ipsilateral side were examined in correlation to the lateral side (Fradet et al. 2010). However, the study of Fradet and colleagues did not examine a hydraulic damping ankle-foot device in correlation to any other type of ankle-foot articulation. To provide a deeper analysis of the effect of MC-AF during slope descent would be beneficial to examine the effects of different types of prosthetic ankle-foot articulation on whole lower-limb system motion. The investigation of different ankle-foot articulations during slope descent is critical as compensatory mechanisms require identification for comfortable and safe ambulation.

2.9 Summary of literature review

To provide a deeper understanding of prosthetic device functionality during slope descent requires investigation of the biomechanics of TTs with different types of prosthetic ankle-foot mechanisms. After lower-limb amputation, the rehabilitation process aims to return amputees to their common daily activities. For lower-limb amputees' day-to-day activities the main consideration is safety. Another consideration in the rehabilitation process is how to reduce the effects of compensatory mechanisms as these could lead to a secondary physical condition. Hence, assessment and analysis of the newly developed lower-limb prosthesis in these tasks are crucial. One of the demanding daily activities for lower-limb amputees is slope descent. Indeed, articulated 'ankle' mechanism improves the biomechanics of slope descent compared to rigid 'ankle' as slope descent requires to increase 'ankle' range-of-motion. Prosthetic 'ankle' mechanisms could have different articulations: elastic (rubber-snubber) (*Epirus*), hydraulic (*Echelon*) (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). These modulated prosthetic devices have replaceable carbon fiber ‘heel’ and ‘fore-foot’ keels (dynamic response). Those ‘ankle’ articulations designed for overground with self-selected walking speed, so would not alter plantar/dorsi-flexion resistance according to the level of ambulation or walking speed. To return active amputees to their daily activities as slope ambulation require prosthetic ankle-foot that could adapt to these activities. Under review in the present study are prosthetic ‘ankles’ which have a modified hinge joint that activates during stance phase according to approach surfaces and walking speed. However, currently even the most advanced prosthetic ‘ankle’ is an oversimplified model of a biological human ankle (Leardini et al. 2000). Prosthetic ‘ankles’ could not fully replicate the complexity of a human ankle. For example, if overground gait amputees walk slower it is partly due to reduced propulsion, ‘push-off’ from the prosthetic foot but during slope descent, this would require an increase of body motion control due to increased gravitational potential energy. Hence, the prosthetic foot would require more braking/ absorption providing.

After trans-tibial amputation patients often compromise the biomechanics of gait in order to retain a safe and efficient gait cycle. Biomechanics of gait were compromised, according to prosthesis functionality and the approaching terrains. Lower-limb amputee biomechanics of gait unlike able-bodied individuals and include asymmetry during the stance time, as to maintain balance on the prosthetic side is more difficult. Propulsion and braking forces with prosthetic foot devices were reduced and as a result, involve the remaining joints to compensate it. It is widely recognised, that lower-limb prosthetic devices with improved functionality could enhance the gait pattern, which could reduce energy cost (Buckley et al. 1997; Hsu et al. 1999; Au et al. 2009). Hence, improved prosthetic ‘ankle’ functionality would reduce biomechanical compensations during slope descent.

The newly developed microprocessor controlled quasi-passive hydraulic ankle foot *Élan* (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) claims to adapt to different terrains and walking velocities. The *Élan* foot has a microprocessor that acts according to speed and slopes ambulation when necessary. The *Élan* device provides independent hydraulic control of plantar/dorsi-flexion it keeps toe up after maximal dorsi-flexion to provide clearance in swing phase and gives the amputee a more natural ankle motion. The *Élan* device would be more appropriate for slope ambulation as it provides adaptive motion compared to rigid or articulated, but non-adaptive prosthetic 'ankle'. It changes the hydraulic 'ankle' damping required to achieve optimal and safe gait pattern for amputees. The main concern of the present study is slope descent. The microprocessor should reduce resistance towards plantar-flexion to provide easier foot-flat followed by an increase in resistance towards dorsi-flexion for safe slope descent. Hence the microprocessor should adapt to walking speed and terrain (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK).

Also, some biomechanical studies analysed gait of lower-limb as an able-bodied mode, but a prosthetic foot cannot be viewed as the human ankle. The research of Prince and Winter (1994) overcame this by using energy-based approach to calculate the scalar power of the prosthetic foot distally (Prince et al. 1994). Recently, Takahashi and colleagues (2012) used this approach; they created a unified deformable segment (*UDS*) model without an evaluation of 'ankle joint' centre (Takahashi et al. 2012). Hence, in order to deliver accurate results in future work, it is important to employ the *UDS* model.

The analysis of literature points to gaps in current knowledge. The investigation of the effects of prosthetic 'ankle' foot designs during the ramp descent on spatio-temporal parameters, joint kinetic and whole body angular momentum is required. Correlation of the effects of microprocessor controlled hydraulic 'ankle' dampening (*Élan*) compared to hydraulic 'ankle' dampening (*Echelon*) or 'rubber-snubber' (*Epirus*) prosthetic 'ankle' during the ramp descent should be investigated. The effect of the restricted ankle in the able-bodied individual should be assessed to deliver the difference in functionality required between ramp descent and overground gait.

2.10 Thesis specific aims and objectives

The main aim of this thesis was to determine the effects of using (adaptive) a microprocessor-controlled hydraulically articulated ankle-foot device in action (MC-AF) mode compared to non-active mode (nonMC-AF) or elastically articulated ankle-foot device (elastic-AF) on the biomechanics of ramp descent in active unilateral trans-tibial amputees (TTs). The comparisons were made between ‘ankle’ articulation mechanisms in prosthetic devices, where prosthetic ankle-foot devices share the same heel and fore-foot carbon fibre keels. To expand understanding of prosthetic ankle-foot articulation an ancillary aim was used to investigate the effects of unilateral restriction of ankle motion on-ramp descent and overground gait in able-bodied participants. Able-bodied participants ankle motion was restricted using a custom made ankle-foot-orthosis (AFO) to simulate an ankle-foot prosthetic device. To achieve the specific objectives of the thesis required:

- Investigation of kinematical adaptations of the whole body motion on the unilateral ankle restriction in able-bodied individuals during ramp descent and overground gait. To expand understanding of kinematical adaptations to a prosthetic ‘ankle’ articulation in TTs.

- Investigation of further kinetical adaptations of remaining joints on the unilateral ankle restriction during overground and ramp descent in able-bodied individuals. To expand further our understanding of kinetic adaptations to a prosthetic ankle-foot articulation in TTs.

- Comparison of the effects of prosthetic ankle-foot articulation on the whole body transition during ramp descent over slow and self-selected walking speeds in TTs.

- Determining how different prosthetic ankle-foot articulations affected contribution of remaining joints during a ramp descent over different walking speeds in TTs. Also, to investigate how the use of the unified deformable segment model quantifies different prosthetic ankle-foot articulations during ramp descent performance.

CHAPTER THREE - GENERAL METHODOLOGY

3.1 Introduction

The chapter contains details of ethical approval, participants with inclusion and exclusion criteria, equipment and methodologies used. Laboratory calibration with the results and justification for the methodologies used regarding scientific literature are included, along with the marker set and biomechanical model used in the study. The specific equipment and methodologies for particular studies are presented within the following chapters

3.2 Ethics

Ethical approval for this study was granted from the University of Bradford's Committee for Ethics in Research (ref. number E.119). All documentation and protocols were conducted in accordance with the tenets of the Declaration of Helsinki. All participants had verbal descriptions and instructions for the test on the day of the test. Prior to data collection, the participants received a copy of and were able to become acquainted with the "Patients information sheet" (Appendix 1). They also signed a "participant consent form" (Appendix 2, 3) and submitted information for a "baseline data proforma" (Appendix 4). Able-bodied volunteer participants were recruited from the students and staff of the University of Bradford by word of mouth and posters. Amputees were recruited from the volunteer group of Alpha and Beta Testing of Blatchford prosthetic devices. To participate in the study able-bodied and amputee participants had to meet the inclusion and exclusion criteria described in section 3.3.1 and 3.3.2.

3.3 Inclusion and exclusion criteria

3.3.1 Inclusion and exclusion criteria of able-bodied participants

The study inclusion criteria requirements were physically active, male participants. The study excluded participants with any known gait impairment or musculoskeletal disorders that may influence locomotion, or currently taking medications for any neurological, cardiovascular, or metabolic disorders, or have a history of lower-limb injuries or surgeries. The age of able-bodied participants does not match the age of amputee participants. The data for able-bodied participants were collected at the beginning of the study; it was not possible to recruit amputees' participants of the same age due to the limited number of amputee participants available who matched the inclusion and exclusion criteria.

3.3.2 Inclusion and exclusion criteria for amputee participants

The study included amputee participants with unilateral below the knee amputation, which have had their amputation for at least two years prior to data collection. Participants were eligible with any residual limb length along with wearing any current or investigated prosthetic devices (*Élan, Echelon, Epirus*; Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). All included participants were able to walk independently and used their prosthesis daily without any self-rated discomfort in their residuum or intact-limb that could interfere with locomotion. The study included Alpha and Beta Testing volunteer participants from Blatchford that classified at least K3 on Medicare Functional Classification Level (MFCL) system by the experienced prosthetist (Chapter 2.4). To be eligible for participation, to confirm amputees' level of activity and exclude amputee's who were unable to perform basic tasks, subjects were screened against a self-report questionnaire the LCI-5 before taking part in the research (The Locomotor Capabilities Index) (Appendix 1) and had to score the maximum 56 points.

The study included participants with limb amputation due to trauma, infection or heart condition (loss of the limb due to congenital heart defect). Subjects with the following conditions have been excluded from the study; neurological; vascular; musculoskeletal disease; pathology affecting balance (subjects self-reported not having dizziness which is likely indicated the absence of a vestibular deficit (Andersson et al. 2006)), sensory dysfunction; peripheral neuropathy and diseases that limit current function as well as taking medication that would affect coordination, balance or reaction time.

3.4 Participants

3.4.1 Able-bodied participants

In the study twenty active males participated (mean (SD) age 27.5 (8.0) years, mass 84.5 (11.5) kg, height 1.79 (0.06) m) (Table 1). Able-bodied participants were recruited for a period of 10 months.

Table 1. Baseline data Performa of able-bodied male participants.

Participants	Age (Years)	Mass (Kg)	Height (M)
PT 1	34.6	89.6	1.81
PT 2	29.2	79.7	1.83
PT 3	20.7	78.3	1.77
PT 4	34.5	82	1.82
PT 5	27.4	89.8	1.86
PT 6	26.3	81.2	1.81
PT 7	28.7	63.4	1.68
PT 8	29.1	96.5	1.83
PT 9	19.7	67.3	1.69
PT 10	24.1	88.1	1.83
PT 11	21.1	70.5	1.76
PT 12	21.1	108.1	1.83
PT 13	26.8	86.7	1.79
PT 14	42.2	82	1.83
PT 15	22.7	72.8	1.75
PT 16	20.9	78.3	1.71
PT 17	21.5	79.7	1.72
PT 18	50.6	93.6	1.74
PT 19	28.5	102.2	1.88
PT 20	20.8	99.2	1.92
Mean (SD)	**27.5 (8.0)**	**84.5 (11.5)**	**1.79 (0.06)**

3.4.2 Amputee participants

Nine physically active, male TT (mean (SD) age 41.2 (12.9) years, height 1.76 (0.06) m, mass 74.14 (15.7) kg, time since amputation at least 2.5 years (mean (SD) 7.5 (6.4), range 2.5 - 22.9 years) prior to participation. All subjects wore the current prosthesis on a daily basis for at least a half a year (mean (SD) 1.6 (1), range 0.5 - 3 years) without any self-rated discomfort during the locomotion. Four amputees habitually used an *Elan* and four *Echelon VT* (Chas. A Blatchford and Sons, Ltd, Basingstoke, UK) and one a *Re-flex Rotate* (Ossur, Reykjavik, Iceland). The prosthesis Re-flex Rotate (Ossur, Reykjavik, Iceland) is non-hydraulic, non-articulated dynamic response feet with the shank-pylon as a spring loaded shock absorber (~2.5 cm of vertical compression) between the foot and the socket. To investigate if participants' having a familiarity with habitual ankle-foot devices (*Elan* or *Echelon VT*) had an effect on results a 'between factor' was used in a mixed-design ANOVA. The participant with a Re-flex Rotate habitual foot was included in the *Echelon VT* group. However, all participants have habitually used or had experience with an articulating ankle-foot device. All amputees wore full-contact sockets which were a custom-made polypropylene thermoplastic. The comfort of the socket is gained by the foam liner and a prosthetic sock. The amputees utilised a cosmetic covering between socket and residuum. Details of the unilateral TTs that participated in the study are presented in table 2. The age of amputee's does not match the age of able-bodied participants. The amputees recruited were aged older than able-bodied participants by 13.7 years. Amputee participants were recruited into the study for a period of 12 months.

Table 2. Baseline data Performa of TT amputee participants.

TT subjects	Age (Years)	Mass (Kg)	Height (M)	Side of Amputation	Time since Amputation (Years)	Habitual Foot device
CL1	50.8	67.1	1.79	Left	22.9	*Elan*
CL 2	39.1	55.8	1.65	Right	2.5	*Elan*
CL 3	42.5	65.4	1.77	Right	8.2	*Elan*
CL 4	22.4	60.8	1.74	Right	3.3	*Echelon VT*
CL 5	63.8	74.6	1.74	Right	2.7	*Re-flex Rotate*
CL 6	24.5	78	1.78	Right	6.1	*Echelon VT*
CL 7	49.8	102	1.74	Right	3.7	*Echelon VT*
CL 8	41.4	67.2	1.78	Right	7.5	*Echelon VT*
CL 9	36.9	96.4	1.86	Right	10.4	*Elan*
Mean (SD)	**41.2 (12.9)**	**74.14 (15.7)**	**1.76 (0.06)**		**7.5 (6.4)**	

3.5 General participant preparation

Prior to tests participants were asked to maintain their usual diet, activity level and refrain from drinking alcohol for 24 hours before the visit. All participants had to wear lycra shorts and a tight top during the experiments. Able-bodied participants had to wear shoes provided for them, and amputee participants had to wear their comfortable flat-soled shoes which they normally use for everyday walking. If the participant has a visual condition and wears corrective spectacles for walking, they were asked to wear them. The data were collected for each participant on the same day. Prior to data collection, participants were measured for height, weight with a habitual prosthesis (for amputees) and clothes. The height of all participants was measured with shoes by using a wall-mounted measuring rod with a single sliding calliper (H-629-1, MARSDEN Weighing Machine Group, Henley-on-Thames, UK). The body weight of the participants was recorded from the force platform (AMTI, MA, USA) in Newtons during stationary standing and converted into kilograms by dividing on $9.81 m/s^2$.

3.6 Laboratory set-up

The data collection was performed in the Biomechanical Laboratory (F9 of Richmond building), University of Bradford. The parameters of the laboratory: width 5.8 m, length 7.0 m and height 2.8 m. To maintain a uniform level of lighting for all tests, the lab had fluorescent tubes, and all windows in the laboratory were covered by black roller blinds. The laboratory is illuminated by six light fittings with each fitting containing three fluorescent tubes. This provided an average illumination of ∽ 400 Lux on the laboratory floor. The room temperature was maintained at ∽ 20°C to provide a comfortable environment for participants. To minimise the risk of distraction, the sign 'EXPERIMENT IN PROGRESS, PLEASE DO NOT ENTER' was on the locked door prior to all data collection. The laboratory floor is raised above the floor level of the main building to make the mounting of the force platforms possible. The floor of the laboratory is covered with green non-slip linoleum (vinyl).

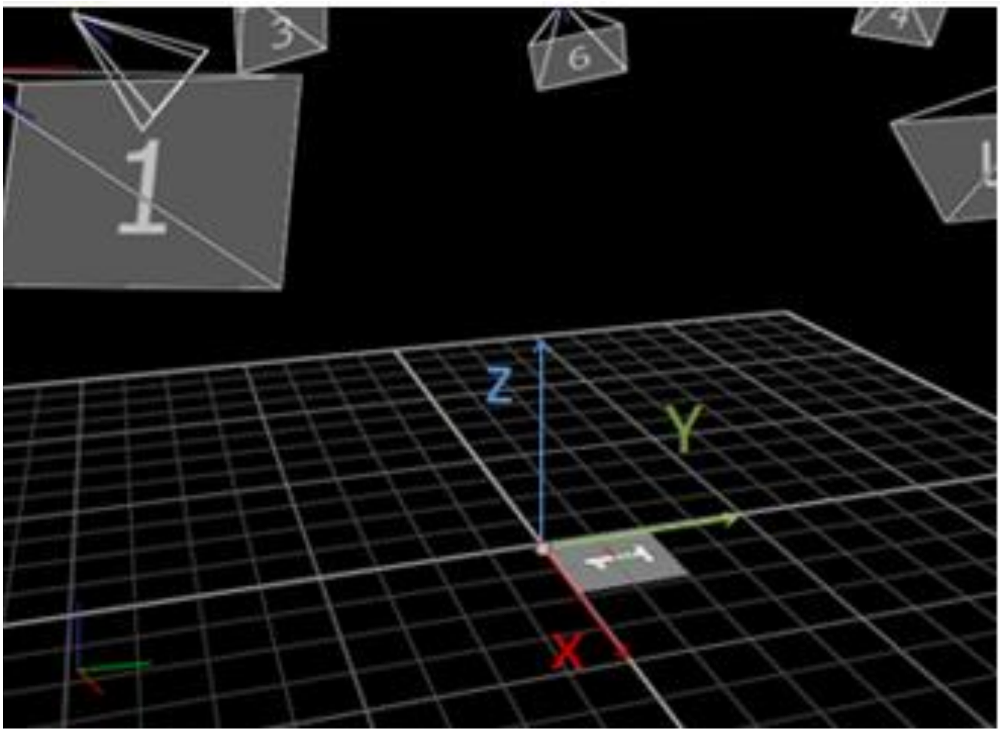

Figure 12. Screen shot from software Vicon Nexus 1.8 (Vicon MX, Oxford, UK). The laboratory visual representation with positive: X (red), Y (green) and Z (red) axis at the origin of the global laboratory coordinate system that is located at the left corner of the force platform.

The origin of the laboratory was specified prior to each data collection. The left (closest to the door force platform as you enter) corner of the force platform is specified as the coordinate origin with the positive X axis towards the short side (464 mm) of the force platform. The positive Y-axis perpendicular X-axis on a ground level towards the long side (508 mm) and direction of travel in overground gait. The positive Z axis is vertical up. The origin of the global laboratory coordinates system (X, Y, Z) is located at the left corner of the force platform and presented in figure 12.

3.7 Experimental equipment

3.7.1 Force platform

The force platform (FP) is a complex electronic device (based on strain gauges). The force transducers acquire kinetic data that quantifies ground reaction forces (GRF) and the centre of pressure (CoP) trajectory. Strain gauges were constantly electrified during functioning. Each strain gauge provides a changed output with a change of electrical resistance under an applied load on the top of an FP. The FP has strain gauge transducers that can provide Fx, Fy, Fz, Mx, My, Mz – channels. Forces are acting towards vertical (Fz), anterior-posterior (Fy) and medial-lateral (Fx) directions. Mx, My, Mz are moments of rotation around X, Y, Z axes with the centre of application in the left corner of FP where are positive moments specified by ‘right-hand rule’ along the axis so anti-clockwise moments are positive. The data accuracy depends on the reliability of several specific parameters: threshold, sensitivity, range, hysteresis, crosstalk and cable disruption. In addition, the manufacturer recommended warming-up the amplifiers for accurate data recording. To ensure correct data overground and ramp decent trials the FP was zeroed, and the amplifier was reset prior to each set of trials.

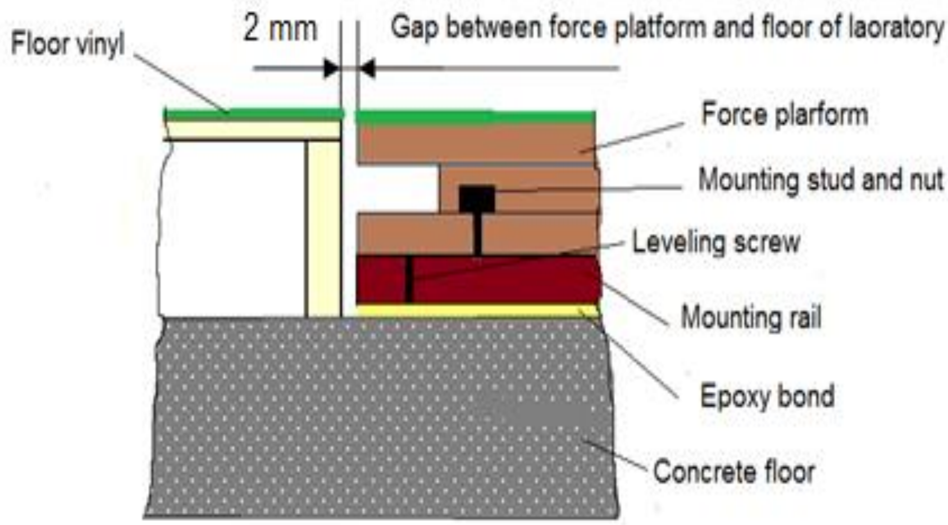

Figure 13. Schematic view of the installed force platform.

In the laboratory, two OR6-7 model Biomechanics Force Platform Model OR6-7-2000 (AMTI, MA, USA) has crosstalk below 2.0% on all channels, Fx, Fy, Fz hysteresis ± 0.2% of full-scale output and Fx, Fy, Fz non-linearity ± 0.2% full-scale output. Typically CoP accuracy ≤0.4 mm and crosstalk values ± 0.2% of applied load with measurement accuracy ± 0.25% of calibration applied load.

(www.amtiweb.com). The FP signal was transmitted through the amplifier MSA-6 (AMTI, MA, USA) and converted from analogue to digital to Dell PC. The sampling rate was set at 400 Hz, which was provided by Vicon Nexus system (VICON, Oxford, UK). The FP was integrated into the floor in the middle of the laboratory to allow participants to achieve the required movement velocity before the force platform and after for deceleration. The dimension of the FP was 464 x 508 x 83 mm (Width x Length x Height) with 2 mm gaps between the FP and the floor (Figure 13). The FP is positioned to the long side towards the direction of travel, so it increases the ability to land 'clean' within the boundaries on the FP if the step length alters.

To reduce systematic errors of the vertical channels in the FP, a quarterly static calibration was performed. The vertical calibration was performed by placing the calibrated weight with a known mass (2, 3, 5, 10, 20, 30, and 50 kg) on the FP in turn and recorded for a few seconds. The calibrated weights with known mass delivered vertical forces (weight ×9.81 m/s^2) when the output compared to the calculated values. To examine the entire surface of the FP, the weight was placed on different points of calibrated FP. The research threshold on the FP was used above 20 N in the vertical direction which corresponds to the threshold set-up in Vicon Nexus 1.8 software (Vicon MX, Oxford, UK). The examination of crosstalk was performed by assessment of outputs from X and Y horizontal channels when only a vertical force was applied on the calibrated FP. The horizontal calibration required specific equipment.

3.7.2 Cameras and calibration equipment

Kinematic data were recorded using a ten-camera infrared system; there were eight cameras MX-3 and two cameras MX-13 (Vicon MX, Oxford, UK) with a sampling rate set at 200 Hz. All cameras were equipped with an infrared strobe. The cameras were mounted on a ceiling, approximately 250 cm above the floor level, to maximise the volume within the accurate motion capture (Figure 14). The volume was in the middle of the laboratory. To record data, software Vicon Nexus 1.8 on Dell computer (Model Precision T 1650 was used, Processor Intel (R) Xeon (R) CPU E3-1290 at 3.70GHz, RAM 8 GB).

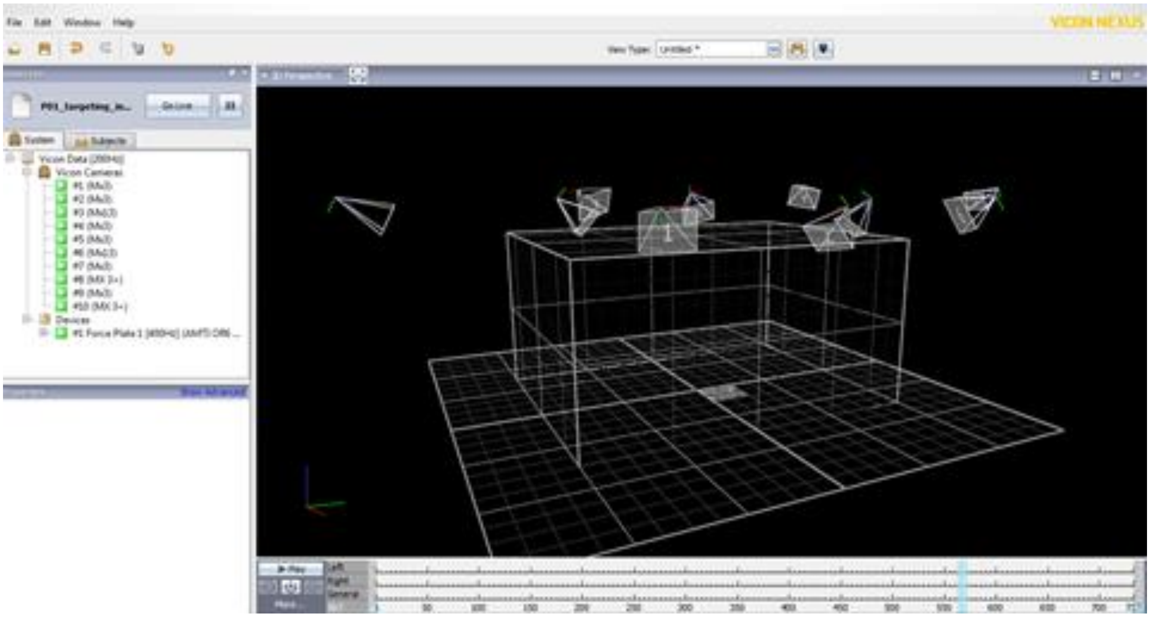

Figure 14. Screen shot of Vicon MX camera position and orientation with the volume from Vicon Nexus 1.8 software (Vicon MX, Oxford, UK).

The calibration procedure was performed prior to each data collection on the volume system with the moving of 3-Marker Wand (390 mm) (VICON, Oxford, UK) which is presented in Figure 15 (1). The orientation of the cameras in 3D perspective within XYZ coordinate origin is set-up by the Clinical L-frame (VICON, Oxford, UK) (Figure 15 (2)) by placing it on the left edge of the FP. The Clinical L-frame was arranged to be parallel with the FP. The camera calibration had error below 0.05 mm in 3D perspective. The positive Y axis was set in the direction of the participant's overground gait travel, according to the International Society of Biomechanics (ISB) recommendations (Wu and Cavanagh 1995).

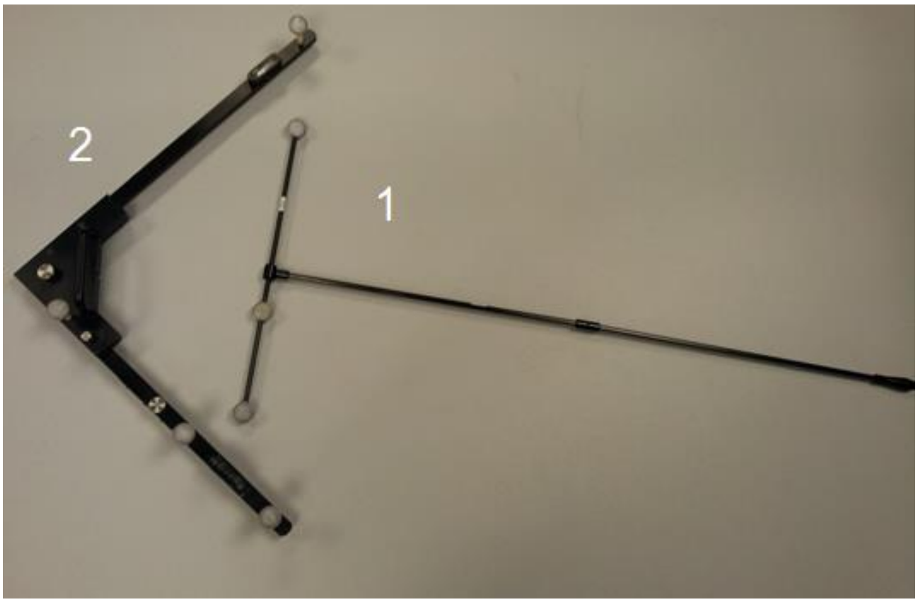

Figure 15. Laboratory set-up equipment. 1- 3 Marker Wand (390 mm); 2- Clinical L-frame.

3.7.3 CalTester

The calibration technique was performed to ensure that the force platform (FP) location and the laboratory motion capture was synchronous and will deliver a correct calculation of participants joint kinetics and Centre-of-Pressure (CoP) location. The *CalTester* (C-Motion, Germantown, MD, USA) is a specifically designed tool which is supplemented by software *CalTesterPlus* (C-Motion Inc., Germantown, MD). However, the research was integrated into Visual3D (v5) software. The tool was utilised to examine the measurement error of the rod orientation and rod tip differences in the three coordinates (x, y, z) of the CoP location (Figure 16, 3). The function is based on examination of the differences between the pointed engineered rod tip in motion capture and the FP data. The FP provides direction and ensures the quantity of action of the force that is applied to the endpoint of the rod. The software examines the error between CoP measurements from the FP and motion capture from cameras as well as the direction and quantity of applied forces (Holden et al. 2003). To ensure the accuracy of all data the examination was performed on the FP and inclined solid block that was designed to transfer the forces (bolted on the top of the FP) to the FP (Figure 16, 1 and 2). The standard examination technique contained five trials in each of the four corners and the middle of the FP. This verified the accuracy of recorded data with spatial synchronisation between the FP, the inclined block and motion capture. Therefore, accurate gait data required an examination of these devices to improve the accuracy of the calculations between the motion capture and FP coordinate systems.

To ensure the accuracy of kinetic data a *CalTester* (C-Motion Inc., Germantown, MD) was used between the Centre of Pressure (CoP) orientation of the force platform and the motion capture within the laboratory reference system (Holden et al. 2003). The rod was applied to five points on the surface of the inclined solid block (four corners and the middle). A ‘force structure’ was constructed in Visual 3D software (C-Motion Inc., Germantown, MD) with the exact dimensions of the inclined solid block. The function of a force structure allows the CoP (x, y, z) coordinates to be transformed within Visual 3D from the platform to the top surface of the ‘force structure’. CalTester assessed he inclined solid block as a 'force structure' in Visual 3D software. The mean error and standard deviation (SD) in force orientation 0.8 (±1.6) degrees between ground reaction forces (GRF) vector and the *CalTester* rod orientation. Mean difference and (SD) in determining CoP x, y, z coordinate measures between the force platform output and the tip of a Cal-tester rod as a motion capture was X=-3 (±3); Y=0 (±3); Z=-2 (±1) mm. The results over five trials are presented in Appendix 5. Any error could be dependent on several factors such as an FP configuration, interface, and alignment of an FP. The main consideration of the research is the alignment of the FP with a removable, inclined block that is used to transfer forces. The removable inclined block, when installed on top of an existing FP, has the 5-degree angle relative to the laboratory coordinate axis.

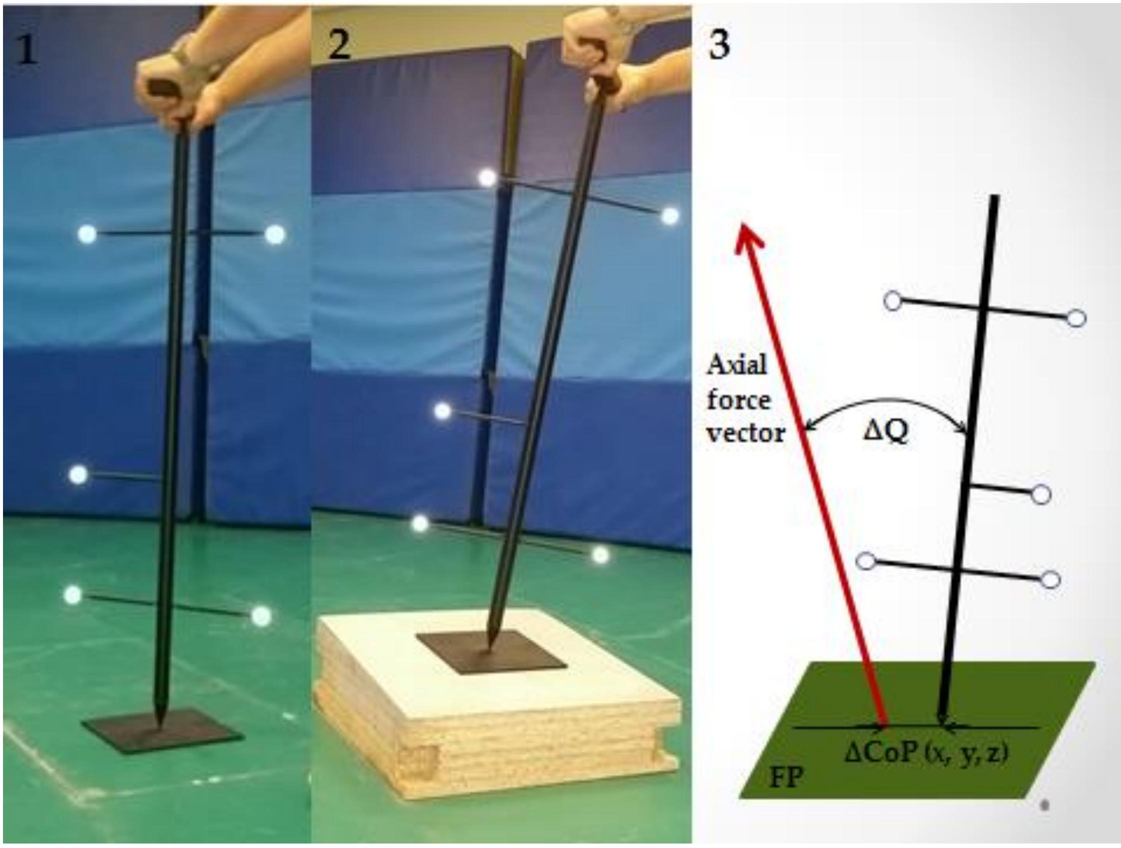

Figure 16. A *CalTester* (C-Motion, Germantown, MD, USA) device is used to apply force to the FP (1) and the inclined block (2); 2 – The differences in force orientation and coordinates the location of the CoP (x, y, z) between *CalTester* rod and axial force vector.

3.7.4 Passive Retroreflective Markers

For a visual representation of participant gait, it is necessary to record the position and translation of segments in virtual 3D space. To create the visualisation, markers are used which can be moved freely within the segment by a participant. This eases the experience of tracking markers in 3D space by a motion capture system. The optical motion capture system has the benefit of a wireless connection, larger volume and good accuracy. The 3D position of passive markers can provide accurate data recording by a motion capture system. However, there are a number of problems which could occur with the motion capture of passive markers: obstruction of the marker, missing marker or even mislabeling between frames. Even in advanced motion capture systems, it would require a significant amount of editing which could also lead to miscalculations. Passive markers are commonly retroreflective spheres that are not actively luminescent, so cameras use a special filter to reflect the light.

Passive retroreflective (PR) markers were used to define anatomical landmarks and track participants body segments in 3D space. In the study markers with a diameter (Ø), 14 mm were used, fixed on a base with a height of 2 mm (19, 2). The motion of the thighs and the shanks were tracked by four anatomically curved thermoplastic clusters 2 mm thick with non-slipping material on a back (2 x thighs and 2 x shank) with four passive PR markers each Ø 14 mm (Figure 17, 3). The position of the head was tracked using a headband with four PR markers Ø 14 mm. (Figure 17, 1).

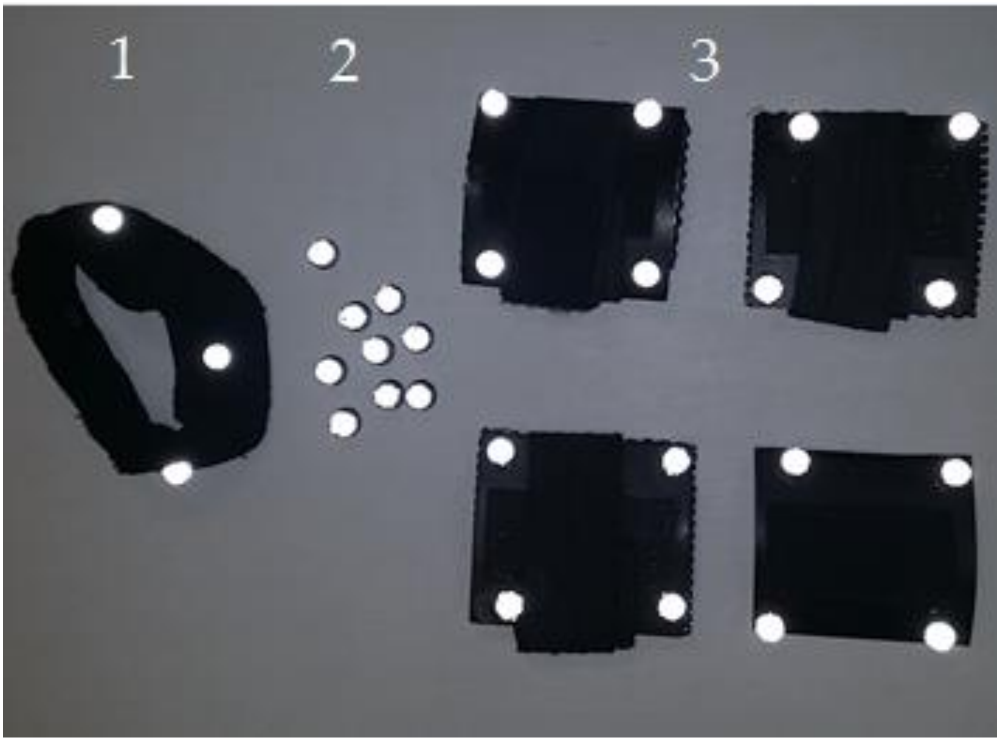

Figure 17 1 - the headband with four Passive Retroreflective (PR) markers diameter (Ø) 14 mm; 2 – PR markers with Ø 14 mm on the base of 2 mm; 3 - four clusters with Ø 14 mm PR markers on anatomically bend thermoplastic.

3.7.5 Digitizing Pointer

A Digitizing Pointer (C-Motion, Germantown, MD, USA) was used in order to create virtual markers in a three-dimensional model without placing actual markers (Figure 18). The Digitizing Pointer (60 cm for large volumes) was used to locate virtual markers on a shoe bottom rim which accurately identified the lowest points of a participant's shoe. Virtual markers (points) were tracked by 3 PRMs of the identified segment. Previously, identifying the border of a shoe was performed by taking direct measurements with a calliper from the middle of the marker to the rim of the shoe. The utilisation of the Digitizing Pointer tool allows the fast and automated creation of virtual markers in Visual3D with further fast and accurate data processing. In this research, virtual markers (points) were created in order to identify points on a shoe tip and a heel which was used to create gait events. The created gait events are described in a further chapter (3.13 Biomechanical data acquisition, processing and analysis). The method would increase the accuracy of the event creation and would reduce the possibility of error.

Figure 18. Digitizing pointer (60 cm for large volumes) (C-Motion, Germantown, MD, USA).

3.7.6 The ramp

The study protocols involve gait on the slope. The modular ramp was designed with a 5 degree of inclination. The ramp gradient was chosen according to the maximum Ramp Gradient in British Standards (BS 8300: 2009) which is 1:12 ~ 5˚. The ramp was designed to enable easy transportation and installation as the biomechanical laboratory is on F floor in the University building. The ramp design also has to provide fast installation. The ramp was built using 12 mm plywood and has five sections. The first (1000x1300x16 mm) and last (1000x1000x261 mm) sections have a level ground surface. Three middle sections have 5 degrees gradient with total length 2800 mm and width 1000 mm. In the middle section of the ramp, a slot is located for the FP block. The middle section of the ramp to ensure alignment and prevent any shifting was fixed to the laboratory floor with four bolts (M10) to pre-installed in the floor nuts (flushed with floor level). In setup condition, to transfer forces without perturbation, the ramp has a constant 2 mm gap with an inclined solid wooden block. The inclined solid wooden block was placed and fixed with two threaded rods (M14) to threaded holes of the existing FP (threaded holes made in accordance with the manufacturer's instructions (AMTI, MA, USA) in order to prevent shifting and ensure alignment of the wooden block with the middle section of the ramp. For fast and consistent alignment relative to the laboratory, the ramp has four points of fixation to the laboratory floor. The inclined solid block also has two points of fixation to the FP to eliminate possible errors due to block movement. The surfaces of the ramp and FP block were painted with grey anti-slip paint (“Toolstation”; Anti-Slip PU Floor Paint Grey code 32762; Hazard safety codes: Xn, R10, R65, R66). The static coefficient of friction measured with a horizontal pull slip metre spring balance (Salter Super Samson, OurWeigh, UK). The anti-slip paint of ramp surface provides a static coefficient of friction equal or greater than 0.62. A similar static coefficient of friction emerges in the study of evolution required coefficient of friction for safe ramp approach (Fino and Lockhart 2014).

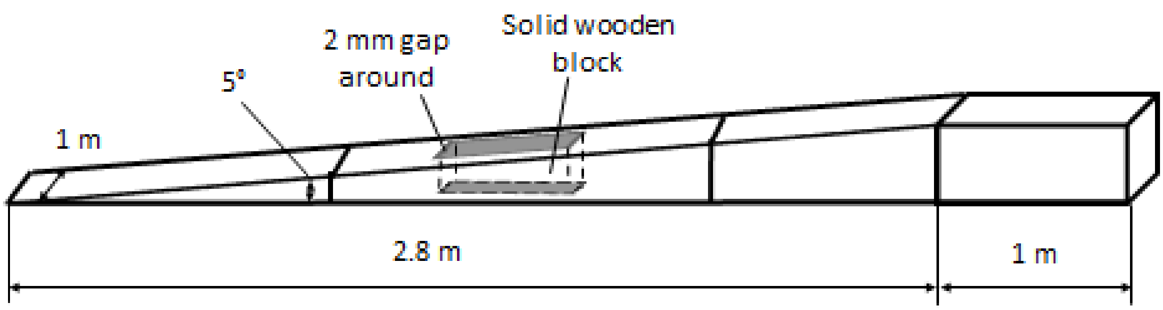

Figure 19. The schematic view of an assembled modulated ramp 2.8 m long (5 degrees of inclination) with 1m of the landing platform.

3.8 Six Degree of Freedom (6DoF) biomechanical model

The six degrees of freedom (6DoF) marker model is one of the most frequently used marker models presently used in the biomechanical community (Buczek et al. 2006; Vanicek et al. 2009; De Asha et al. 2013b). To create a biomechanical model a 6DoF marker set with 54 markers was used. The model contained nine segments: feet, shank, thighs (left and right); pelvis; trunk and head (Cappozzo et al. 1995). To create a 6DoF Visual 3D software (Visual 3D Professional v5.00.21) was used. The 6DoF model refers to the free motion of a segment in 3D space. Each segment has three variables that specify the location of the origin and the other three variables specify the rotation of each of the principal axes of the segment. To identify each segment requires three or more markers. To reduce the total number of markers, some segments shared markers and define a joint centre. The pelvis segment is defined using the iliac crest (proximal) and greater trochanter (distal) markers where the length of the segment was the distance between the iliac crest and trochanter markers. The sacrum markers used to track a segment.

Human anthropometry has a critical influence on the prosthetic design and efficiency. Visual3D software (C-motion Inc., Germantown, MD) used to calculate a segment mass based on Dempster's regression equations (Dempster 1955). Dempster's regression equations were further updated according to Zatsiorsky-Seluyanov's segment inertia parameters by de Leva (de Leva 1996). The alternative evaluation of anthropometrical body segments parameters was presented in the Technical Report (1166.03) of Contini and Drills in 1966 (Drillis and Contini 1966), which updated parameters still widely employed by the biomechanical community.

The attachment of markers to soft tissue does not provide accurate data on motion due to soft tissue artefact between the marker and the bone (Leardini et al. 2005). Certainly, direct attachment of a marker to the bone will provide more accurate results, but the use of invasive marker placement is not always feasible (Fuller et al. 1997), and medical ethics would conflict with that. In this study, non-invasive placement of the optimal markers was chosen. Consequently, the investigation of non-invasive marker placement method during gait has indicated intersubject similarity of soft tissue artefacts (Gao and Zheng 2008). Nevertheless, the error of the soft tissue artefact can be reduced by the placement of markers on a stationary part of a joint (Karlsson and Tranberg 1999). The joint markers were removed after dynamic trials due to the higher possibility of 'losing' this marker during locomotion. To reduce the error of soft tissue artefact clusters were used (Cappozzo et al. 1997; Manal et al. 2000). The markers are fixed on the non-flexible surface so do not move relative to each other. In this study, clusters were used to identify four segments bilaterally: shank and thigh. The markers on the clusters (tracking markers) are located to define the distal and proximal ends of a tracked segment. Each segment has a coordinate system with the centre of rotation at the proximal endpoints. The centre of foot origin is an ankle joint, the centre of a shank is a knee joint and origin of a thigh is a hip joint. To evaluate anatomical joint centres, a functional joint centre technique was employed (Schwartz and Rozumalski 2005). Functional joint centres were defined on all anatomical lower-limb joints (bilaterally: hip, knee, ankle), including the ankle with custom AFO. The technique is based on the functional joint centre of two rigid segments can be defined by the least point of rotation between segments (Greenwood 1988). The joint midpoint between segments defines the joint centre between these segments. The segments are defined on proximal and distal endpoints by anatomical markers. The midpoint between the two distal landmarks was defined as the distal endpoint of a segment. A segment local coordinate system is defined at the proximal joint centre. Z-axis is along the line that joins distal to proximal endpoints of a segment with positive value towards proximal endpoints. Y-axis is perpendicular to the Z axis in the frontal plane. The x-axis is perpendicular to Z and Y through the medial and lateral direction by the right-hand rule. The head segment does not have a proximal joint (endpoint), so the origin was defined as the midpoint of the two posterior landmarks. The four border targets in the frontal plane are used to compute the least square fit in Visual 3D software (C-Motion, Germantown, MD, USA). The least squares were computed from the sum of squares distance between four targets, and the frontal plane was minimised. This specified the targets to identify the distal and proximal end of a segment which will affect the location and orientation of a Segment Coordinate System in Visual 3D software (C-Motion, Germantown, MD, USA).

In prosthetic devices, the distal endpoints of the shank are not necessarily an articulation point, so the use of the standard inverse dynamic approach would not be appropriate. Therefore, the model prosthesis feet allow for proximal and distal endpoints of a segment to motion relative to each other as an energy flow due to deformation of the heel and fore-foot keels (Prince et al. 1994). The deformation of the heel and fore-foot keel provided the simulation of plantar/dorsi-flexion motion and modelled without an evaluation centre of rotation in ankle joint (Takahashi et al. 2012) as it would appear in an inverse dynamic approach. To normalise the distal endpoint of the pylon/shank between prosthetic feet as a result of 'ankle' absence was created on the shank's mid-line at the same height as a functional joint centre of the contralateral ankle.

The whole body CoM model velocity has good agreements with a CoM velocity of lower-limb and trunk segments within valid accuracy (Vanrenterghem et al. 2010). Therefore, it was unnecessary to add upper limb segments to be used for the whole body CoM model representation. To reduce data collection and processing time the whole body CoM model was represented only by lower-limb and trunk segments.

3.9 Locations of Passive Retroreflective Markers

The laboratory volume calibration and set up the origin of the laboratory, described in section 3.7.2 (Cameras and calibration equipment), was followed by recording a static calibration file of a participant. The static calibration file requires only one frame within all attached passive retroreflective (PR) markers. The static calibration of the participant was detected in 3D perspective by the Vicon Nexus system (Vicon MX, Oxford, UK). To identify participants anatomical landmarks and the sacrum PR markers were used which were attached to the participant by double-sided tape. Double-sided tape was placed on the bottom of a PR marker base. Segments of the participant were identified by clusters which were attached by elastic Velcro straps. The placement of PR markers was performed according to the participant's anatomy with attachment to the skin or clothing. The study utilised nine segments 6DoF model with head, trunk, pelvis and bilaterally (left/right): thighs, shanks and feet. PR markers and clusters were placed on the participant according to figure 20. The marker set was labelled in Vicon Nexus 1.8 software (Vicon MX, Oxford, UK) according to the list in Appendix 6.

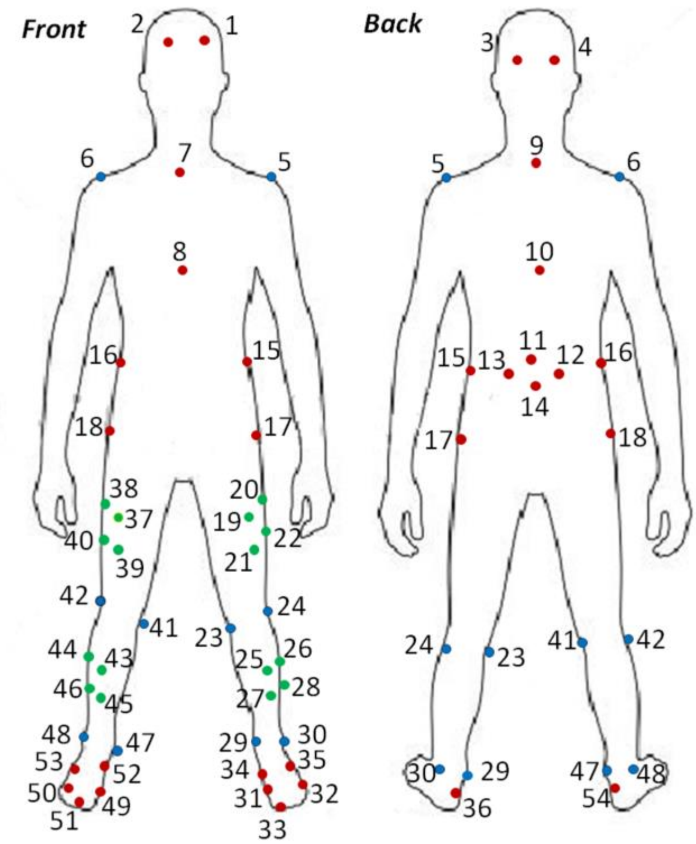

Figure 20. 6DoF marker set (front and back contour) was used. Red dots - anatomical landmark PR markers; green dots – clusters; blue dots – calibration PR markers. The list of numbers, labels and position of markers is presented in Appendix 6.

PR markers (red dots) were placed bilaterally (left; right) on the anatomical body landmarks (or equivalent locations on the prosthesis): iliac crest directly above the greater trochanter (15; 16), greater trochanter (17; 18), cluster of four markers was placed the sacrum (11;12;13;14), superior aspects of first and fifth metatarsal heads (31, 49; 32, 50), distal end of second toe (33; 51), pragmatically on the medial and lateral aspects of the mid-foot (34, 52; 35,53) and posterior calcaneous (36; 54). Markers were also placed on the sternal notch (7), xiphoid process (8), vertebrae C7 (9) and T8 (10). A head band was used to mount four head PR markers that located to define left/right and anterior/posterior position of the head. The clusters were anatomically curved plates with four mounted PR markers. The clusters were worn on the thighs and shanks, while four PR markers were attached to skin or clothing about the sacrum. Tracking PR markers (red and green dots) are used to compute the motion. Calibration PR markers (blue dots) were placed bilaterally (left; right): acromion process (5; 6), medial and lateral femoral condyles (23, 41; 24,42), medial and lateral malleoli (29, 47; 30,48).

Labelling and gap filling of marker trajectories were undertaken within Vicon Nexus 1.8 software (Vicon, Oxford, UK). The fill pattern of a missing marker according to a marker with similar motion and by spline fill. That is an automatic method in Vicon Nexus software extrapolates trajectories based on the known motion of the marker. Labelled C3D files were exported to Visual 3D motion analysis software (C-motion, Germantown, MD, USA), where nine segments 6DoF model of each participant was constructed. The functional joint centre technique was employed to evaluate anatomical joint centres (bilaterally: hip, knee, ankle) (Schwartz and Rozumalski 2005). The technique was supported within the Visual 3D software.

3.10 Static and dynamic calibration files

After accurately placing PR markers, clusters and a headband to ensure accurate representation of the model recorded a static calibration file, the subsequent phase was to record a static calibration file correctly. The assessed participants were required to stand still in the anatomical position for three seconds. The static calibration file of this anatomical position was used as the reference position to determine the segment embedded axes of segments and joint angles between segments. All PR markers have to be seen by the cameras, during recording the static calibration file, to allow assignment with the Visual 3D model template (C-Motion, Germantown, MD, USA).

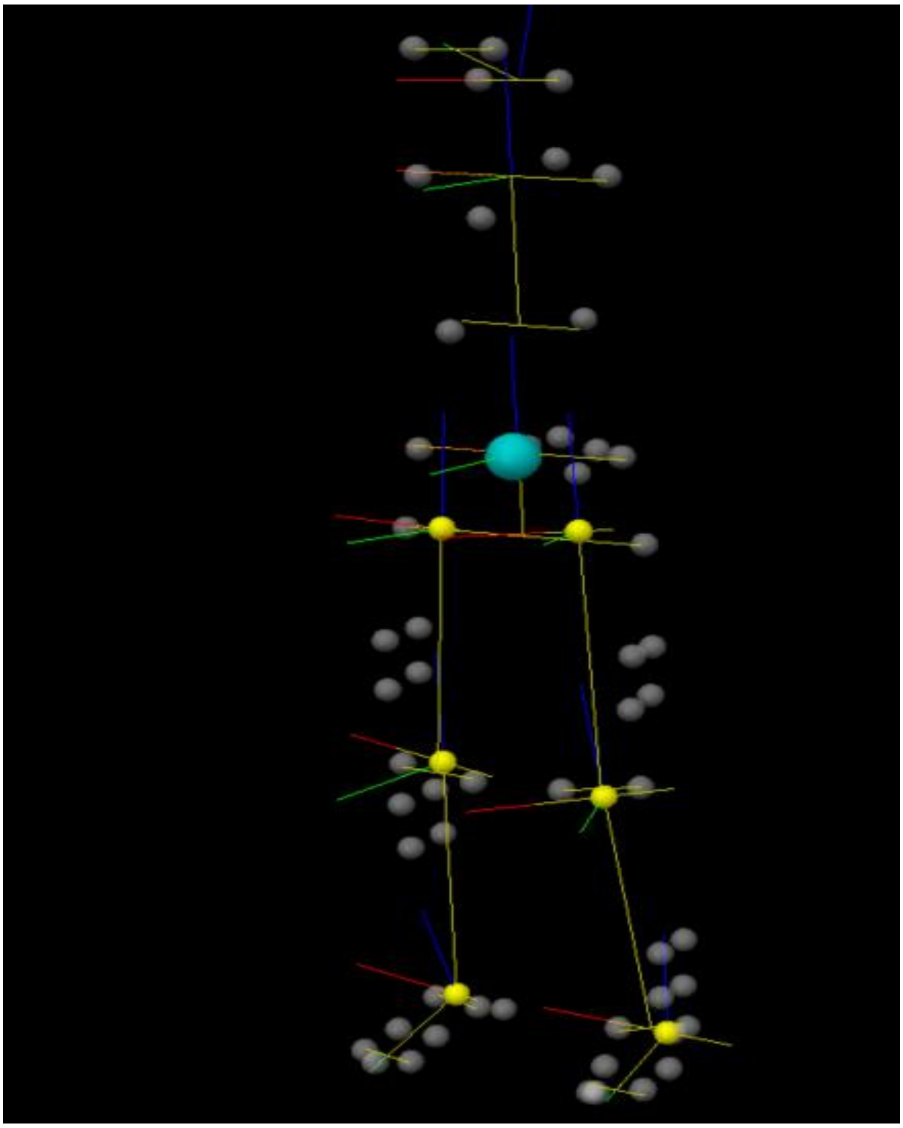

Figure 21. Screen shot presenting 6 DoF model of static calibration file from Visual 3D (v5, C-Motion, Germantown, MD, USA). Grey dots represent PR markers, Yellow dots with the local coordinate system represent joint centres. Blue ball is a Centre of the Mass location of the participant.

After ‘subject’ static calibration, the calibration PR markers were removed after recording successful static calibration file to eliminate distortion of the participant's gait due to its location and risk being knocked off during dynamic trials. Calibration PR markers are used to define the segments. To know the position of a segment requires a minimum of three markers so the additional number of markers would reduce the risk of ‘losing’ a segment with occlusion of one of the markers. Based on this, each segment of the model used a minimum of four PR markers. To track a foot segment, six PR markers were used so it would significantly reduce the risk of ‘losing’ a segment during data collection.

Following the ‘subject’ static calibration procedure all temporary calibration PR markers were removed. To calculate lower-limb joint angles, the proximal segment used as the reference segment. The hip angle is the angle between the thigh and pelvis. The knee angle is the angle between the shank and thigh. The neutral ankle angle is the angle between the virtual foot and shank. In the static calibration trial, the ankle angle position was used to create a virtual foot for all biological ankles. The evaluations of functional joint centres (FJCs) movements of the lower-limb joints were performed. The participants were required to perform left and right limb movements of the hip, knee and ankle to identify the FJCs. The data were recorded for each lower-limb joint, where motions were performed. The hip joint movement was done in the directions of flexion/extension, adduction/abduction and circles clock/anti-clock wise for 3-4 second each. The knee joint movement was done in the direction of flexion and extension for 10 seconds. The ankle FJC was evaluated from the motion in plantar-flexion and dorsi-flexion for 10 seconds. This technique is based on the research of Schwartz and Rozumalski, is currently the most accurate and has advantages compared to previously used methodologies (Davis et al. 1991; O'Brien et al. 1999). To assess FJC of the intact side in TTs, amputee participants had to be weight bearing on the prosthetic side which was challenging due to balance difficulty on the prosthetic side (Hermodsson et al. 1994; Hof et al. 2007). To perform the required joint motion, amputee participants used a stabilising pole to support themselves. The evaluation of FJC for residual knee was critical because the knee joint location has a significant influence on this joint moments (Holden and Stanhope 1998). Prosthetic socket fit examined by an experienced prosthetist to ensure sufficient residual knee flexion. To palpate and locate the femoral condyles markers within the socket of the prosthesis was challenging as the joint was enclosed within the socket. The motion of the prosthetic ‘ankle’ device was excluded from the FJC method as deformation of carbon fiber heel and fore-foot keel incorporated with prosthetic ‘ankle’ device so does not have a constant joint centre, due to the difference of applied load and approached terrains. To examine prosthetic feet power generation and absorption, a foot model defined as ‘unified deformable segment’ was used (UDS) (Takahashi et al. 2012; Takahashi and Stanhope 2013). A one prosthetic foot model was used throughout this thesis.

3.11 Ankle Foot Orthosis and walking protocol

All able-bodied participants, in this thesis, were wearing custom made ankle-foot orthotics (AFO) (Figure 22). To investigate the effects of ankle articulation the AFO was designed with a hinge that allows manipulation of the ankle range of motion in the sagittal plane. The AFO was used to restrict sagittal plane ankle motion of the right limb. The AFO’s medial and lateral struts each had a lockable hinge located approximately at ankle height. The AFO construction has two modes: restricted (‘locked’) and unrestricted (‘unlocked’). The modes were controlled by two grub screws either side of each hinge. In ‘unlocked’ mode the hinge allowed total range of motion 30 degrees in plantar/dorsi-flexion direction. For the restricted mode the AFO’s hinges were fixed (‘locked’) by screwing down the two grub-screws either side. The AFO’s in restricted mode did not fully immobilise ankle motion in the sagittal plane and allowed the motion of approximately ±3-5 degrees in plantar/dorsi-flexion direction, which were estimated between all able bodied participants. The AFO’s restricted mode was performed with the hinge that was ‘locked’ by screwing down two grub-screws either side of the hinge on each strut when a participant was standing in an upright position. The proximal end of the AFO struts was fastened with two Velcro straps around the right shank and adjustments were made to each participant while they are standing in the right position to ensure the alignment of the hinges was as close as possible to the axis of each participant’s anatomical ankle. The distal ends of AFO struts were inserted in the heel channel to ensure alignment as close as possible to the axis of the participant’s anatomical ankle. Participants wore the same model of flat sole shoes (The UK size 8-11) that were designed for AFO with an integrated fixation (small channel cut into the raised heel) in which the distal ends of the AFO's struts were inserted.

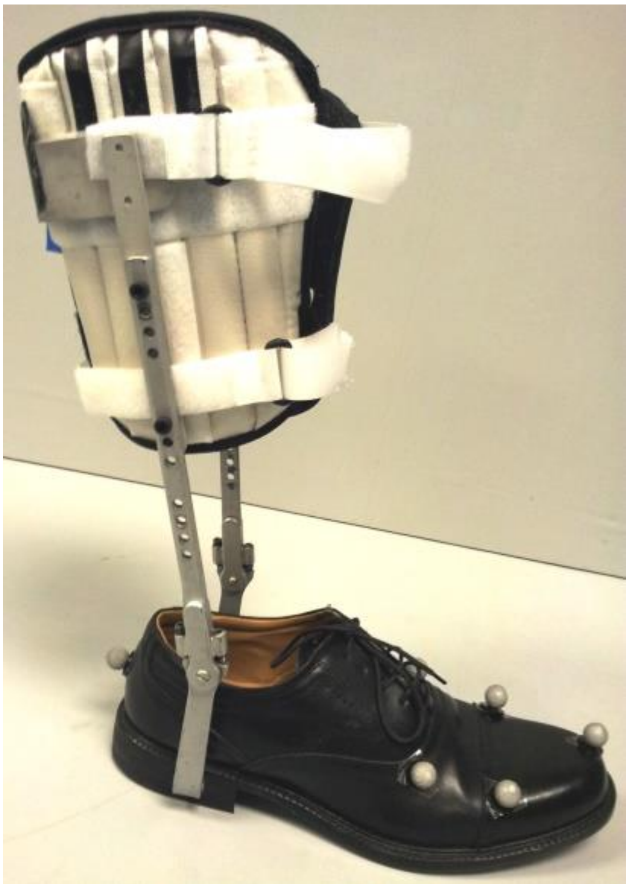

Figure 22. Custom made ankle-foot orthosis (AFO).

In able-bodied participants, the PR markers were placed on lower-limbs bilateral. Nevertheless, on the involved side, positions of markers have been acknowledged below. To avoid additional calculations between shank and AFO motion did not require an estimation of the brace (unloading) moment separate to the ankle moment (Schmalz et al. 2010). The cluster was placed directly on the shank distally, to track involved side shank segment. The evaluation of the ankle joint centre on AFO side was employed FJC method. To define ankle FJC, calibration PR markers were placed on medial and lateral malleoli. The design of custom AFO allowed the cluster and calibration markers to be placed according to the 6 DoF model. The data collection for each able-bodied participant was performed in one session. Prior to testing, each participant was fitted with a custom made AFO. All participants and had familiarisation with the laboratory by level ground walking and with typically two familiarisation trials prior data recording for each block (overground and downslope). Prior to each trial, participants were instructed as to which limb they should initiate gait.

Throughout the session, each able-bodied participant completed two blocks of repeated gait trials, one involving walking down the ramp and the other along the level ground (i.e. laboratory floor without a ramp). Participants were instructed: to walk at a normal, comfortable speed, i.e. at their freely chosen speed across the laboratory. Block order was counterbalanced across participants. The counterbalance of block order helps to negate any learning or fatigue effects, so half of the participants (10) have been randomly allocated to start walking either overground or ramp descent first. Each block included two ankle conditions, restricted ('locked') and non-restricted ('unlocked') the order of which was also randomly counterbalanced across participants. The walking speed was not controlled as controlled speed could affect typical walking pattern within effect intra-subject variability (Shiavi et al. 1987) as a result of walking pattern modification according to require walking speed. Starting location was adjusted for each participant to ensure the involved limb landed 'clean' within the force platform boundaries without gait adjustments. The participants were asked to descend the ramp (Figure 23) in a controlled self-selected manner. Each participant completed 6 successful trials for the right (involved) and left (non-involved) foot with AFO in each ankle mode. The total number of trials completed was 48: 6 (repetitions) x 2 ankle conditions (non-restricted and restricted), x 2 limbs (involved (with AFO) and non-involved (contralateral), x 2 gradients (overground and downslope).

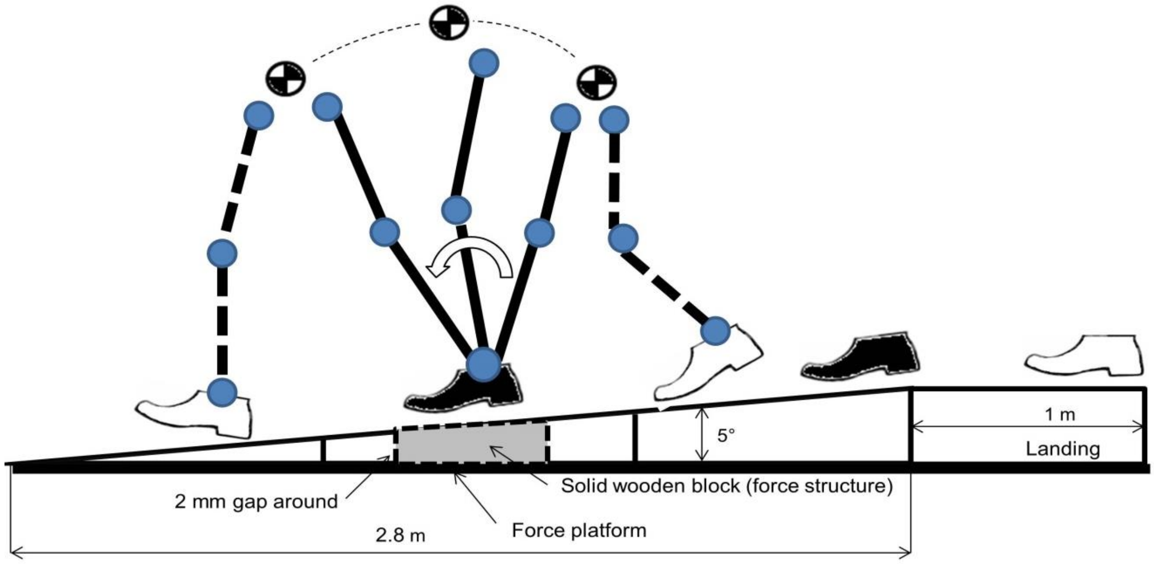

Figure 23. Modular ramp (2.8 metres) has 5˚ of inclination and landing 1 metre. The third step landed on the wooden block. The integrated wooden block on the top of force is transfer forces.

3.12 Prosthetics and protocol

The data collection of amputee participants was acquired on the same day and was split into two blocks for assessment. One used the elastically articulated prosthetic ankle-foot device *Epirus* (elastic-AF) (Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) (Figure 24 A). Another was a quasi-passive microprocessor-controlled hydraulic ankle-foot device *Elan* (Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) (Figure 24 B) with uniaxial articulating motion. The *Elan* device examined in two modes: active (MC-AF) and non-active (nonMC-AF). The *Elan* device in non-active mode behaves just like the *Echelon* (Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) hydraulic ankle-foot device. The manipulation between active and non-active modes of the *Elan* device was performed remotely by the prosthetist via Bluetooth connection to the device.

The *Elan* device in the nonMC-AF mode perform as the *Echelon* device and has default settings that control the rates of articulation and is considered to be optimal for the participant’s self-selected walking speed in overground gait. Default settings vary across amputee participants and depend on a combination of the participant feedback regarding perceived comfort with function at optimal for the participant's self-selected walking speed. The participants with the habitual *Elan* device used their habitual default settings. *Echelon* users used settings which were set by an experienced prosthetist. The *Elan* device responds when users approach inclined surfaces by changing the rates of articulation in plantar/dorsi-flexion direction The maximum ankle-foot range of motion of 6° and 3˚ for plantar-flexion and dorsi-flexion respectively from 'neutral' standing position. The damping settings have the range from 1-9 and independently control plantar-flexion and dorsi-flexion motion. The microprocessor controls these settings by altering the position of a valve in plantar-flexion and dorsi-flexion motion. The *Elan's* hydraulic damping settings were optimised by the fine tune of plantar/dorsi-flexion damping according to a combination of the prosthetist experience and participant feedback about comfort and function during the familiarisation period. In the ramp descent mode, the plantar-flexion resistance goes to its second lowest setting, and the dorsi-flexion resistance goes to its second highest. To aid the reader, the *Elan* in active mode can be termed as being in *Elan,* MC-AF or adaptable mode. The *Elan* in non-active mode can be termed as nonMC-AF. *Elan* and *Epirus* (elastic-AF) prosthetic devices which were equipped with an independent heel (to absorb shock during heel contact and return energy during mid-stance) and split toe, fore-foot leaf-spring keels (to assist during push-off) that provide tripod stability. The deformation properties of carbon fiber springs provide a good energy response for dynamic response foot (DRF) device users. These devices were manufactured by Chas. A. Blatchford and Sons Ltd., Basingstoke, UK.

Prior to data collection, all amputee participants were examined by the same qualified, licensed and experienced prosthetist. All amputee participants were familiar with articulated ankle-foot devices. Either ankle-foot device: *Elan* or *Epirus* (Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) was altered by exchanging the existing prosthetic foot for each participant. Swapping ankle-foot devices was performed by the same prosthetist according to the specific requirements of the amputee and prosthetic device used and kept as near to constant as possible. A prosthetist set up and aligned the prosthetic ankle-foot devices for optimal use in overground level with self-selected walking speed.

The pylon (shank) alignment, suspension, socket and the total length of the prosthesis were kept as near as possible across the assessed ankle-foot prosthetic devices (as were the size and stiffness of the heel and fore-foot keels). The pylon length with *Elan* device has to be shorter due to the raised building height compared to the *Epirus* elastically (rubber-snubber) articulated device, so the length of the pylon was adjusted (shortened or replaced with suitable longer length). All adjustments were performed in order to achieve optimal alignment to have an adequate comparison between prosthetic devices. The heel and fore-foot keels (size and stiffness) were kept the same across assessed prosthetic devices within the subject. The prescription of the prosthetics and its components and settings vary between subjects according to weight, age, activity level, cost, and preference and gait specifics. All amputees in this thesis wore a custom-fitted full contact thermoplastic socket to provide an interface between residuum and the prosthesis. All participants were familiarised with each type of ankle-foot device by walking on the level floor of the laboratory for approximately 20 minutes. Prior to recording kinematic and kinetic data, participants were allowed to familiarise themselves with the walking tasks and 'practice' typically two trials, data collection protocols.

The difference between the *Elan* (1.2 kg) and the *Echelon* (0.9 kg) is only 300 grammes. Nevertheless, the study examined the stance phase only, so inertial properties are irrelevant and not assessed at this time. The *Epirus* (elastic-AF) has a spherical rubber-snubber ankle device that provides multi-axial articulation between the pylon and heel/fore-foot keel section. The ankle-foot device can 'plantar-flex' up to 15° but not 'dorsi-flex', as 'dorsi-flexion' is restricted by a 'hard stop' within the mechanism (Figure 24 A). After data collection amputees' were given back the habitual prosthesis within a habitual setting that was returned to its original condition.

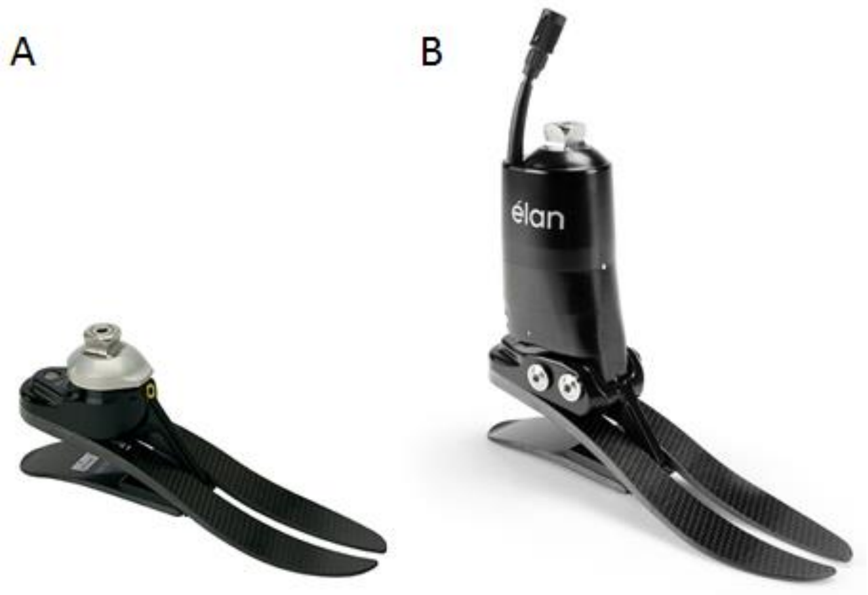

Figure 24. (A) -*Epirus* (Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) and (B) -*Elan* (Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) ankle-foot prosthetic devices were used in the study (adapted from www.blatchford.co.uk; accessed 20/05/2016).

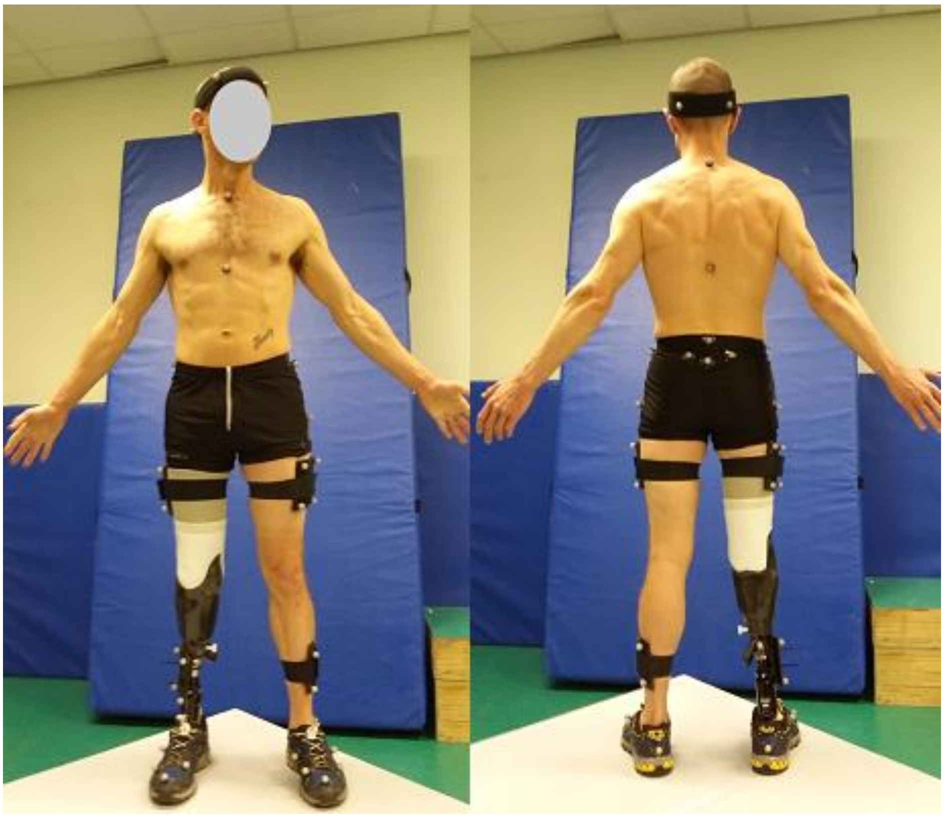

Figure 25. TT participant with *Elan* (Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) and 6 DoF marker set (front and rear view) prior data collection. The calibration PR markers that define joint centres (left/right knee and ankle) and left/right acromion were removed.

All amputee participants had 20 minutes to become familiar and comfortable with the laboratory and each prosthetic foot (*Epirus* and *Elan*). Prior to data collection amputees had a typically two trials to familiarise with the task. Participants were instructed as to which limb they should lead. Prosthetic feet (*Epirus* and *Elan*) order was counterbalanced across participants. The *Elan* 'ON' (MC-AF) and 'OFF' (nonMC-AF) modes were 'blind' for amputees in randomly counterbalanced order (performed in a Microsoft Excel (Microsoft Office Home and Student 2010, Version 14.0.7159.5000). The walking speed was not controlled as a pre-set walking speed could affect the typical gait pattern with intra-subject variability (Shiavi et al. 1987) because participants have to alter the gait pattern according to the required criteria. The TT participants were asked to descend the ramp (Figure 23) with self-selected and comfortable slow walking speeds. The participants were instructed to walk as they would normally walk and comfortable slow prior to these trials respectively. Participants were instructed to descend the ramp, at two speeds: where the first performed was always self-selected and second slow. Each participant completed six successful trials (the third step was landing 'clean' on the force platform without gait adjustments) for three ankle-foot articulation conditions (MC-AF, nonMC-AF and elastic-AF) during those conditions and at two walking speeds (self-selected, slow). The limited number of trials (n=6) and conditions were chosen to avoid data collection becoming a fitness test, so any fine tuning in *Elan* performance could be missed. The reduced numbers of trials could provide the possibility to assess a wider range of TT which would help to deliver more accurate analysis. The total number of trials completed was 72: 2 limbs (prosthetic, intact) x 6 (repetitions) x 3 'ankle' types (MC-AF, nonMC-AF and elastic-AF), x 2 walking speeds (ramp descent self-selected and comfortable slow).

3.13 Biomechanical data acquisition, processing and analysis

The study motion capture (kinematic) of data were collected at 200 Hz using a ten camera Vicon MX system (Vicon MX, Oxford, UK). The system allowed capture of the three-dimensional (3D) motion data via PR markers that were placed on the assessed participant. Kinetic data were collected from a floor-mounted force platform at 400Hz (AMTI, Watertown, MA, USA). Kinematic and kinetic data were recorded on Vicon Nexus 1.8 software (Vicon, Oxford, UK) where labelling and gap filling of marker trajectories was completed. To fill the gap of any missing marker, a marker with similar trajectory or spline fill was used. For further processing, all labelled C3D files were exported to Visual 3D motion analysis software (C-Motion, Germantown, MD, USA). When C3D files transferred to Visual3D software, the force platform frequency data were adjusted to the motion capture frequency data of 200Hz as it was the lowest frequency of collected data. Although nine parameters applied on the FP: Force (F_x, F_y, F_z), Center of Pressure (COP_x, COP_y, COP_z), Free Moments (M_x, M_y, M_z), the FP provides only six parameters of the human motion. These six components are Ground Reaction Force (GRF) (GRF_x;GRF_y; GRF_z), Center of Pressure (COP_x, COP_y), and a Free Moment (M_z). Other parameters: Free Moments M_x and M_y are assumed to be zero and COP_z assumed to be on the top surface of the FP. To transfer the forces and COP from the FP onto the top surface of the inclined solid block surface a 'force structure' was constructed with the dimensions of the inclined solid block used in the study in Visual 3D software (C-Motion Inc., Germantown, MD). To define the 'force structure' parameters implemented, a vertical offset to the surface of the FP where the corner of the 'force structure' was specified. Therefore, the inclined block represents the elevation of the FP corners. The 'force structure' each corner X, Y and Z coordinates were determined, and the implementation does not affect the FP parameters. Where a 'force structure' was considered as a mechanism that combines an FP data and the inclined solid block which is attached to the FP. Visual 3D software transfer COP (x, y, z) coordinates from the platform to the top surface of the 'force structure'. If Visual3D appropriately modified the intersection from the original force vector and COP signals, parameters presented would be coincident and collinear at the place of application (www.c-motion.com/v3dwiki). To validate 'force structure' parameters employed CalTester (C-Motion Inc., Germantown, MD) (Chapter 3.7.3).

A nine segment 6DoF model (Cappozzo et al. 1995) was created, the model constructed includes head, thorax/abdomen, pelvis and lower-limbs (left and right: thighs, shanks, feet) for each participant. To evaluate joint centres in lower-limbs, a functional joint centre method was used (Schwartz and Rozumalski 2005) which is described in section 3.9. Joint kinetics of biological lower-limbs were calculated using the standard inverse dynamics approach and determined by the assumption that segments of the system are rigid and the ground reaction forces acting on the distal end of the segment (Dumas et al. 2009).

The prosthetic foot device does not function as the human ankle so cannot be examined the same. To assess prosthetic foot power was used a unified deformable segment' (*UDS*) model without evaluation of 'ankle joint' centre approach (Takahashi et al. 2012; Takahashi and Stanhope 2013). The proposed *UDS* model can be used for all structures distal to the knee. This eliminated the requirements to model a shank and foot autonomously and allowed the calculation of the scalar power flow during stance. However, this approach can provide scalar power, so unable to differentiate between planes. The prosthetic foot provides the energy absorbed and returned, which is power flow at the distal end of the shank pylon regardless of the type of attachment and/or foot, is the physical application point of the forces and moments transferred to and from the shank (Prince et al. 1994). The distal energy (P_{dist}) at the prosthetic distal end is the sum of translational power (P_{trans}) and rotational power (P_{rot}) in the sagittal plane which is determined by the energy flows. The energy flow is calculated by integrating the power with respect to time during the stance time. The distal energy leaving the pylon is negative and entering the pylon energy is positive (De Asha et al. 2013b).

$$P_{dist} = P_{transl} + P_{rotat} \qquad \text{Equation 5}$$

Translational power (P_{trans}) was calculated as:

$$P_{transl} = (F_z * V_z) + (F_y * V_y) \qquad \text{Equation 6}$$

F_y and F_z are the anterior-posterior and vertical components of the reaction force (N) acting at the pylon distal end.

V_y and V_z are the anterior-posterior and vertical velocities ($m*s^{-1}$) of the pylon distal end.

Rotational power (P_{rotat}) was calculated as:

$$P_{rotat} = M_x \times \omega_s \qquad \text{Equation 7}$$

M - a moment in the sagittal plane that applied at the distal end of the pylon (N*m) and ω_s - angular velocity of the assessed shank segment ($rad*s^{-1}$).

Moment specifies which muscles are active flexors or extensors and by how much. However, moments do not explain why this is happening. Moments parameters are calculated in Newton meters (N*m), in order to normalise this data between participants it requires dividing the weight (N*m/kg) of the participant.

Kinematic data were filtered with a fourth order, zero-lag Butterworth filter with a 6 Hz cut-off according to the recommendation of study Robertson and Dowling (Robertson and Dowling 2003). The recorded data requires filtering as result of marker motion due to skin artefact, electronic noise in optical devices, the error of digitising process and other noises that could affect the data to achieve smooth and accurate data. GRF data were filtered with a 4th order zero-lag Butterworth low-pass filter and the cut-off frequency of 20 Hz.

The sequence of a gait cycle is described below to ensure clarity throughout further analysis of human gait. The gait cycle is divided into stance and swing phase. The key concern of this study is the stance phase. The stance phase of human gait can be divided into three phases: Weight Acceptance, Single-Limb-Support, and Swing Limb Advancement (Perry et al. 1992).

The stance phase of the involved limb was defined from Initial contact (IC) to toe off (TO) when vertical ground reaction forces went above or below threshold 20N, respectively. First or Initial Double support (DS1; 1st rocker) was defined from the involved limb IC to TO_{con} of the contralateral limb. Single-limb-support (SLS) was defined from the contralateral limb TO_{con} to ICcon. Second or terminal double support (DS2; 2nd rocker) was defined from contralateral IC_{con} to TO of the involved limb which also is the end of the stance phase. To define IC and TO of contralateral limb, there was no force platform so kinematic data were used. TOcon events were created according to Zeni gait event detection, which is the first peak in the Y direction of the toe marker relative to the pelvis (Zeni et al. 2008). IC_{con} events were determined by when the heel marker vertical velocity drops below the threshold 0.15 m/s. Joint work and moments were normalised to participants' body weight.

The prior statistical analysis averaged across trials at each condition for each participant was used to provide mean value. All data were examined for normality of the distribution with Shapiro-Wilk Test. The statistical analysis was undertaken in Statistica v6.0 (StatSoft, Inc., Tulsa, OK, USA). To examine normally distributed parametric data repeated measures of variance were used (ANOVA) with comparisons between prosthetics (*Epirus*, *Elan* in active and non-active modes) or ankle condition (restricted, non-restricted) and speed level (slow and self-selected) or inclination level (overground, slope descent) as factors. The specifics of each statistical model are described within the methodology of related studies. Post-hoc analyses comparisons were conducted using Tukey HSD tests. The level of significance was set at p=0. 05, unless noted otherwise. The effect sizes were calculated for each independent biomechanical parameter to quantify the magnitude of the differences between two means on a unitless standard scale. The effect size measure of Cohen's is a distribution-based method that obtained and used to quantify the differences in gait parameters between conditions.

Effect size (d) was regarded as ‘small’ if d < 0.3, ‘medium’ if d = 0.3–0.5, and ‘large’ if d > 0.5. The formula of effect size was presented by Cohen 1992:

$$d = \frac{M_1 - M_2}{\mathrm{SD}_{pooled}}$$

Equation 8

Where d, a measure of effect size; M_1 and M_2, the means at baseline and follow-up accordingly; SD_{pooled} pooled standard deviation at baseline from the cohort (Cohen 1992).

CHAPTER FOUR - ADAPTATIONS OF WHOLE BODY MOTION TO THE RESTRICTION OF THE ANKLE JOINT DURING RAMP DESCENT IN ABLE-BODIED INDIVIDUALS

4.1 Introduction

The functionality of current prosthetic foot devices is limited compared to the human ankle. To have a better understanding of the effects of prosthetic foot device's functionality on gait biomechanics would be beneficial to investigate the effects of the human ankle with restricted motion. The investigation of able bodied participants while wearing a custom made ankle-foot orthosis (AFO) could aid further understanding biomechanics of gait and development of prosthetic devices.

AFO is an external device that is designed to set the foot–shank angle throughout the gait cycle. The objective of a rigid AFO that restricts active plantar/dorsi-flexion throughout the swing phase and as an aid to control shank foot alignment throughout the stance phase (Jaivin et al. 1992). Throughout the swing phase, AFO facilitates the patient's ability to maintain relative dorsi-flexion and so controls foot drag on the ground whilst ensuring safe foot clearance (Perry et al. 1992). The stance phase AFO helps to control foot placement after initial contact with further support of medio-lateral stability throughout the stance phase (Perry et al. 1992; Nolan et al. 2009). Most studies analysed gait with AFO in patients with gait disorders, but a limited number of studies were performed without the gait disorder. To understand the fundamental effect of prosthetic on amputees' gait, it was of interest to know how the restriction of the ankle would affect individuals without the gait disorder.

In healthy individuals ankle articulation contributes to gait pattern throughout the swing and stance phases, so elimination of its functionality would mean it has to be compensated for by the remaining joints (Perry 1974; Lehmann et al. 1985; Lehmann et al. 1986; Balmaseda et al. 1988; Tuggy and Ong 2000). The number of researchers has determined that the restriction of the ankle motion reduces walking speed, step time, step length and stride in overground gait (Murray et al. 1984; Opara et al. 1985; Carlson et al. 1997; Romkes and Schweizer 2015). However, the information presented in the scientific literature on the effects of ankle restriction on the downslope in individuals without gait disorders was not known.

Daily activities involve ambulation on non-level surfaces (McIntosh et al. 2006) and involve adaptation of the lower-limb motion in order to approach inclined surfaces (Lay et al. 2006). Walking upslope requires more energy expenditure due to an increase of work to raise the Centre-of-Mass (CoM) against gravity (Lay et al. 2006; DeVita et al. 2008). Walking down slope has the inverse effect as it involves the kinetic energy growth lowering CoM due to gravity (Chapman 2008). Humans adapt gait pattern for slope descent by reducing the velocity and stride length compared to overground gait in order to control walking speed within kinetic energy growth during slope descent (Kawamura et al. 1991; Sun et al. 1996; McIntosh et al. 2006). To adapt gait pattern requires the contribution of ankle function throughout stance and swing phases (Kuster et al. 1995). In slope descent, during loading response, an ankle plantar-flexion involves adaptation by an increase in braking ground reaction forces (GRF) and power absorption to control CoM gravitational forces (Redfern and DiPasquale 1997; McIntosh et al. 2006). In downslope gait, plantar-flexion in the first phase of the gait cycle is utilised to achieve foot-flat quicker. This adaptation is employed in order to control the CoM gravitational potential energy growth within the forward velocity at the weight acceptance phase and increased requirements to maintain anterior-posterior balance. The active plantar-flexion at the end of the stance phase (push off) is reduced due to unnecessary power generation during the slope descent (McIntosh et al. 2006). The restriction of ankle motion reduces walking velocity during ground level surface, but the influence on gait velocity during slope is unknown.

The gait cycle is divided into stance and swing phases. In a normal gait cycle, the stance phase could be divided into three sub sequential rockers (Perry et al. 1992). The initial double support (DS1) or the first rocker; a single-limb-support (SLS) or second rocker; the terminal double support (DS2) or third rocker. The

SLS phase was used an inverted pendulum (IP) model to explain bipedal gait efficiency (Gage et al. 2004; Hof et al. 2005; Kuo et al. 2005). IP transition over the fulcrum (ankle) requires the energy of the contralateral side. Commonly, researchers up to the present day have linked CoM and Centre-of-Pressure (CoP) to merge both variables (Pai and Patton 1997; Buczek et al. 2006). However, due to the CoP sensitivity during first and third rockers, the use of this variable can be difficult. Elimination of unwanted variance would help to understand the effect of ankle function on whole body motion.

The calibrated volume of the laboratory did not permit data recording of two strides (full gait cycle), so walking speed during stance phase was used in the study. The initial set up of camera positions (cameras were fixed to the celling) and orientation were made to record overground data. To minimise error for overground and ramp descent gait; the camera orientations were adjusted, which affected the calibrated volume of recorded data. A comparative measurement of walking speed employed a new variable, Virtual Limb (VL) which is the angle between support, ankle functional joint centre (FJC) which was evaluated in Visual 3D (C-Motion, Germantown, MD, USA) and linked to CoM. The VL motion is directly related to cadence and step length. The VL would display changes of BW movement relative to the support foot during a stance phase with a repercussion of the ankle motion. The VL angular velocity is directly related to cadence and step length so an increase in angular velocity would be dependent on those parameters. The introduction of a new variable VL would display changes of BW arched trajectory movement relative to the support foot during a stance phase without an effect of the CoP progression and more accurate behaviour of CoM relative to the support foot. To achieve efficient gait, the ankle has to contribute into non-faltering, ‘roll over’ BW transition. Consequently, the restriction of the ankle has to affect the IP model motion, and its behaviour would alter according to the conditions: slope descent and the ankle restriction. Little is known about the effects of restricted ankle motion on slope descent on whole body motion.

To assess the problems regarding the use of a solid AFO during overground and downslope gait, a custom made AFO that could have two modes was utilised: restricted and non-restricted plantar/dorsi-flexion motion. It is important to examine the impact of rigid AFO, on joint kinematics and whole body motion during ramp descent due to the growing demand of AFO utilisation. The present study, examine the effects of restricted ankle motion on the dynamics of body weight transition for overground and ramp descent in able-bodied individuals. The purpose of the study was to investigate the effects of unilaterally restricted ankle motion on the sagittal plane body kinematics and the temporal-spatial parameters during ramp descent and distinguish from overground gait in healthy adult individuals. The ankle articulation during stance phase on an incline surfaces has a substantial role in maintaining dynamic stability (Vickers et al. 2008), so body motion relative to the support foot is crucial. Ramp descent compared to overground gait has a higher risk of loss of balance with a fall or slip in contrast to overground gait (Redfern and DiPasquale 1997). To assess the effect of ramp descent on balance a new VL variable could be employed. VL angular velocity is directly related to gait velocity and step length. The assessment of VL variable during SLS would provide a deeper understanding of full body CoM motion in relation to dynamic stability as a consequence of ankle motion. The hypothesis of the study was that body weight (BW) dynamics during ramp descent would be increased with the restricted ankle as a result of the inability of the ankle to plantar-flex and to control CoM motion against gravity, which would increase VL angular velocity. Alternatively, it would increase the support shank angular velocity with the increase of knee flexion under a loading response. Additionally, to control VL motion during ramp descent gait, step length would be reduced. The restricted ankle would also reduce the step length as a result of limited push off from the braced ankle.

4.2 Methods

4.2.1 Participants and Ethics

Twenty physically active, males (mean (SD) age 27.5 (8.0) years, mass 84.5 (11.5) kg, height 1.79 (0.06) m), participated in this study. Full details are presented in Chapter 3.4.1. Ethical approval for this study was conducted in accordance with the tenets of the Declaration of Helsinki and granted from the University of Bradford's Committee for Ethics in Research.

4.2.2 Specific equipment, procedure, data acquisition and processing

To manipulate ankle articulation, the custom made ankle-foot orthosis (AFO) device was utilised. Details of the AFO and walking procedures are presented in Chapter 3.11.

To examine ramp descent, a custom made modular ramp 2.8 metres long, with an inclination of 5 degrees and 1.0-metre long level ground landing was used. Full details are provided in chapter 3.7.6. Prior data collection and when the solid incline block was removed or installed on the force plate (FP), to ensure correct data the FP was zeroed in Vicon Nexus 1.8 software (Vicon, Oxford, UK) and amplifier was reset. Ten Vicon MX cameras used to capture motion during the overground and ramp descent walking trials (Chapter 3.7.2). The cameras were positioned in a circle surrounding the FP, which is located the centre of the laboratory, to minimise reconstruction error within a calibrated volume. The calibrated volume had dimensions of approximately 3 m (length) x 2 m (width) x 2.5 m (height) (Figure 19). The volume was calibrated prior each data collection using the Marker Wand (390 mm) where Clinical L-frame (Figure 15) used to set up the orientation of the cameras in 3D perspective within XYZ coordinate origin as described in chapter 3.7.2. The volume calibration and the coordinate origin set up were performed without the ramp.

Details of all laboratory equipment used, how kinetic and kinematic data were recorded with the full protocol procedure are presented in chapter three. Participants were identified by passive retro-reflective markers according to six degrees of freedom (6 DoF) model as described in chapter three.

Participants were introduced to the laboratory and prior to recording kinematic and kinetic data, allowed to familiarise themselves with the walking tasks and 'practice' typically two trials, according to data collection protocol (Chapter 3.11). Each participant completed six successful trials (landing precisely on of the force platform without gait alterations) with the involved and non-involved limbs in restricted/non-restricted modes in counterbalance order between the participants. Gait was assessed in two blocks: on a ground level 8 metre long walkway and on the ramp descent that was described in chapter 3.7.6. Participants were instructed to walk at the self-selected walking speed as they would normally walk during overground and down the ramp gait. Prior to each trial, participants were instructed, which limb they should lead with depending on an assessed limb. Overground and ramp descent gait was performed in counterbalanced order among the participants.

The detailed description of data recording, processing and filtering provided in chapter three.

The stance phase was defined from initial contact (IC) till toe-off (TO) and verified from vertical components (Z) of ground reaction forces with a threshold of 20 N. The single-limb-support (SLS) was defined through kinematic data with the stance phase from TO till the IC of the contralateral foot. TO was created according to Zeni gait event detection (Zeni Jr et al. 2008) and IC was determined as the instants where the heel marker's vertical velocity reduced to 0.15 m/s.

4.2.3 Data analysis

The variables were determined during the stance phase on the right (involved) and left (non-involved) limb for each trial and then averaged across the trials to provide the main parameter for each condition. The following variables listed below were assessed: knee flexion loading response (deg.); CoM mean (A-P) velocity was determined during the stance phase by the CoM mean velocity between IC and TO of the involved limb; VL angular velocity at contralateral TO and IC ($^{\circ}sec^{-1}$); shank angular velocity at contralateral TO and IC ($^{\circ}sec^{-1}$); VL mean angular velocity during SLS ($^{\circ}sec^{-1}$); VL standard deviation angular velocity SLS ($^{\circ}sec^{-1}$); VL and shank mean angular velocity during DS1 and DS2 phases ($^{\circ}sec^{-1}$); step length, stance time. VL and shank angular velocity during DS1, SLS and DS2 were determined as the rate of change of angular position of a rotating segment within the global coordinate system in the sagittal plane ($^{\circ}sec^{-1}$). Knee loading response: determined as peak knee flexion following initial contact.

4.2.4 Statistics

To determine differences between restricted/non-restricted ankle conditions and overground/ ramp descent repeated measures in ANOVA were used. The effect of size differences (low $d < 0.3$, moderate $0.3 < d < 0.5$ and high $d < 0.5$) were calculated as Cohen's (Cohen 1988). Statistical analyses were performed in Statistica (v6, StatSoft, Inc., Tulsa, OK, USA). To identify any significance between conditions a post hoc comparison with Turkey HSD tests was used. The level of significance set was $p < 0.05$.

4.3 Results

4.3.1 Involved limb (right limb with AFO)

The involved limb joints angular displacements during overground and ramp descent with restricted and non-restricted conditions are illustrated in figure 28. Figure 29 illustrates the mean values of shank angular velocity, VL angular velocity, and VL length. The mean (±SD) and statistical significance of CoM velocity, VL and shank angular velocity during the stance on an involved (right) limb with AFO in non-restricted and restricted conditions in overground and ramp descent is presented in table 4. Table 5 illustrates the parameters: step length, stance time. Mean loading response peak knee flexion is illustrated in Figure 27. There were no significant interactions between the level of ambulation and ankle condition, so this will not be presented further in the results section unless stated otherwise.

Shank mean angular velocity during SLS led to a reduction to ankle restriction (by ~2-5 °. s^{-1}, $p < 0.001$) and was higher for ramp descent compared to overground gait (trend only, $p = 0.06$). VL angular velocity at contralateral TO led to a reduction of ankle restriction (˚sec-^{1}) (by ~3-4°, $p < 0.001$) and reduced in ramp descent compared to overground gait (by ~2-3%, $p < 0.024$) (Figure 29). VL angular velocity at contralateral IC had no effect unrestricted ankle ($p = 0.33$) but increased during ramp descent compared to overground gait ($p = 0.036$). VL angular velocity SLS mean (˚sec^{-1}) had no effect on ankle condition ($p = 0.11$) but was greater for ramp descent compared to overground gait ($p = 0.046$). VL angular velocity SLS SD (˚sec^{-1}) was reduced with restricted ankle condition ($p = 0.008$) but without the effect of the level ($p = 0.15$). The VL angle at contralateral IC had no effect of the restricted ankle ($p = 0.75$) but increased during ramp descent compare to overground gait ($p < 0.001$). The VL angle at contralateral TO was reduced with a restricted ankle ($p < 0.001$) and reduced during ramp descent compared to overground gait ($p < 0.01$). Shank angular velocity at contralateral TO (˚sec^{-1}) was increased for ramp descent compared to overground ($p < 0.001$) but no effect of ankle condition ($p = 0.63$).

Shank angular velocity at contralateral IC (˚sec-[1]) was increased ramp descent compared to overground gait ($p<0.01$) but reduced with restricted ankle condition (trend only, $p = 0.06$).

Results indicated that participants were (CoM velocity) walking slower on the ramp descent than overground ($p = 0.004$) but restricted ankle did not change this fact ($p = 0.20$) (Table 5). Step length was reduced for the restricted ankle condition, compared to non-restricted ($p = 0.01$) and was reduced for ramp descent compared to overground gait ($p < 0.001$). Stance time was increased for the restricted ankle condition, compared to non-restricted (trend, $p = 0.08$) but had no effect in ramp descent ($p = 0.17$). Loading response knee flexion was increased for restricted, compared to non-restricted ankle conditions (by ~5-6% or ~1-2°, $p < 0.001$) and was greater for downslope compared to overground gait (by ~1-2% or ~5-6°, $p < 0.001$) (Figure 27). Attainment of foot-flat was delayed for restricted compared to non-restricted ankle condition ($p < 0.001$) but was unchanged across surface conditions ($p = 0.12$); there was no interaction between terms ($p = 0.86$). The timing of heel off was unaffected by ankle restriction ($p = 0.35$) or by the surface condition ($p = 0.09$), but there was an interaction between terms ($p = 0.03$). The timing of heel off was delayed for restricted compared to non-restricted ankle condition but only during overground gait.

Table 3 Group mean (±SD) of involved (right) side: Timing to Foot-Flat (sec) Timing of heel off (sec) with AFO in non-restricted and restricted conditions in overground and ramp descent. Where differences between ankle conditions are significant effect sizes Cohen's (d) are presented (in *italics*).

	Overground		**Ramp descent**		
	Non-restricted	*Restricted*	*Non-restricted*	*Restricted*	p value
Time to attain Foot-Flat (sec)	0.14 (0.02)	0.16 (0.03) *0.7*	0.14 (0.02)	0.15 (0.02) *0.8*	level 0.13 **cond.<0.001** Int. 0.86
Timing of heel off (secF	0.495 (0.055)	0.505 (0.059) *0.2*	0.515 (0.055)	0.513 (0.070) *<0.1*	level 0.09 cond. 0.35 **Int. 0.03**

Table 4. Group mean (±SD) of involved (right) side VL angles at contralateral IC and TO; VL and shank angular velocity at contralateral TO and IC; for the period of the single-limb-support (SLS) with AFO in non-restricted and restricted conditions in overground and ramp descent. Where differences between ankle conditions are significant effect sizes Cohen's (d) are presented (in *italics*).

	Overground		**Ramp descent**		p value
	Non-restricted	*Restricted*	*Non-restricted*	*Restricted*	
VL angle at contralateral TO (°)	-8.5 (1.9)	-8.3 (1.9) *0.3*	-6.1 (1.6)	-5.6 (1.6) *0.3*	**Level<0.001** **cond.<0.01** Int. 0.10
VL angle at contralateral IC (°)	21.1 (2.5)	21.0 (2.5) *0.1*	23.5 (2.7)	23.6 (2.9) *<0.1*	**Level<0.001** cond. 0.75 Int. 0.47
VL angular velocity at contralateral TO (°sec^{-1})	81.8 (9.6)	79.3 (9.4) *0.3*	78.4 (10.4)	76.8 (10.4) *0.2*	**Level 0.024** **cond.<0.001** Int. 0.23
VL angular velocity at contralateral IC (°sec^{-1})	69.3 (11.2)	69.1 (8.9) *<0.1*	72.6 (10.0)	71.2 (9.7) *0.1*	**Level 0.036** cond.0.33 Int. 0.44
Shank angular velocity at contralateral TO (°sec^{-1})	90.6 (21.1)	92.9 (21.0) *0.1*	120.0 (26.9)	119.5 (26.9) *<0.1*	**Level<0.001** cond.0.63 Int. 0.45
Shank angular velocity at contralateral IC (°sec^{-1})	150.9 (21.2)	146.4 (26.7) *0.2*	169.2 (33.8)	-165.3 (30.3) *0.1*	**Level 0.002** cond.0.06 Int. 0.85
Shank angular velocity SLS mean (°sec^{-1})	70.8 (11.9)	66.1 (14.2) *0.6*	72.6 (12.7)	70.1 (14.5) *0.3*	Level 0.06 **cond.<0.001** Int. 0.08
VL angular velocity SLS mean (°sec^{-1})	72.7 (9.6)	71.9 (10.0) *0.1*	70.5 (9.7)	70.0 (10.4) *0.1*	**Level 0.046** cond.0.11 Int. 0.66
VL angular velocity SLS standard deviation (°sec^{-1})	4.9 (1.3)	4.5 (0.9) *0.4*	4.2 (1.2)	4.1 (1.3) *0.1*	Level 0.15 **cond.0.008** Int. 0.23

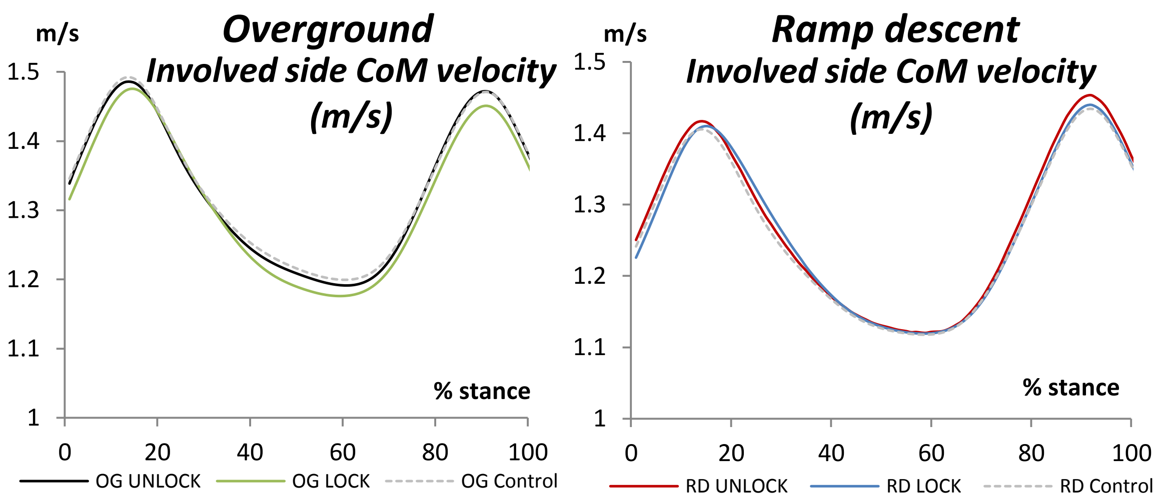

Figure 26 Mean CoM velocity in the anterior-posterior direction and normalised to 100 points (stance phase) and averaged across 20 participants. (OG UNLOCK –overground non-restricted; OG LOCK –overground restricted; RD UNLOCK –ramp descent non-restricted; RD LOCK –ramp descent restricted). OG Control – (dashed grey line) overground control; RD Control – (dashed grey line) ramp descent control. NB for some of the figures the data for the different limbs appears not to be visible (included). This is because the anterior-posterior CoM velocity profile is very similar to another limb condition.

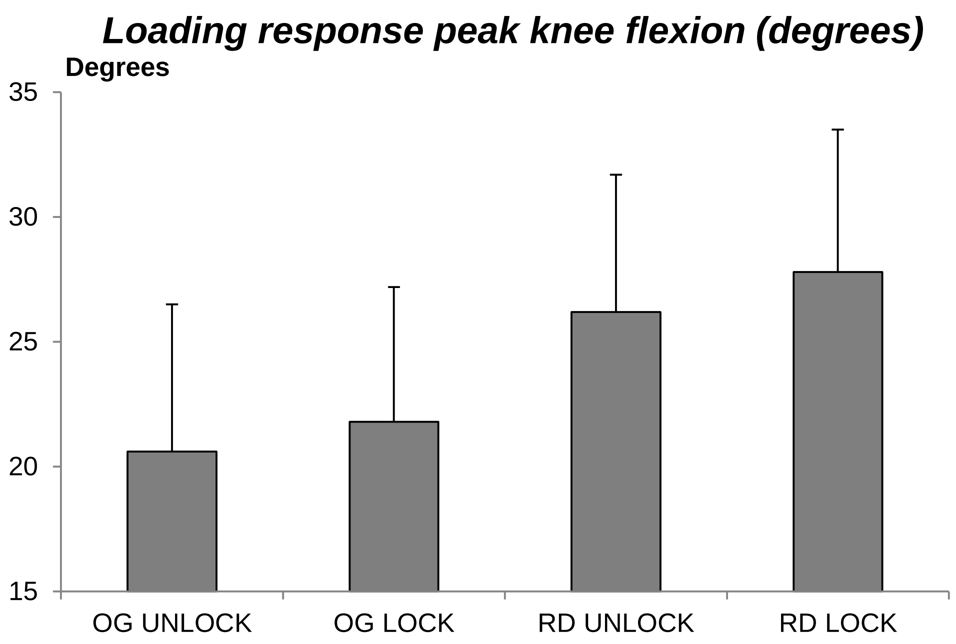

Figure 27 Involved (right) side mean loading response peak knee flexion; ensemble averaged across 20 subjects. (OG UNLOCK –overground non-restricted; OG LOCK –overground restricted; RD UNLOCK –ramp descent non-restricted; RD LOCK –ramp descent restricted).

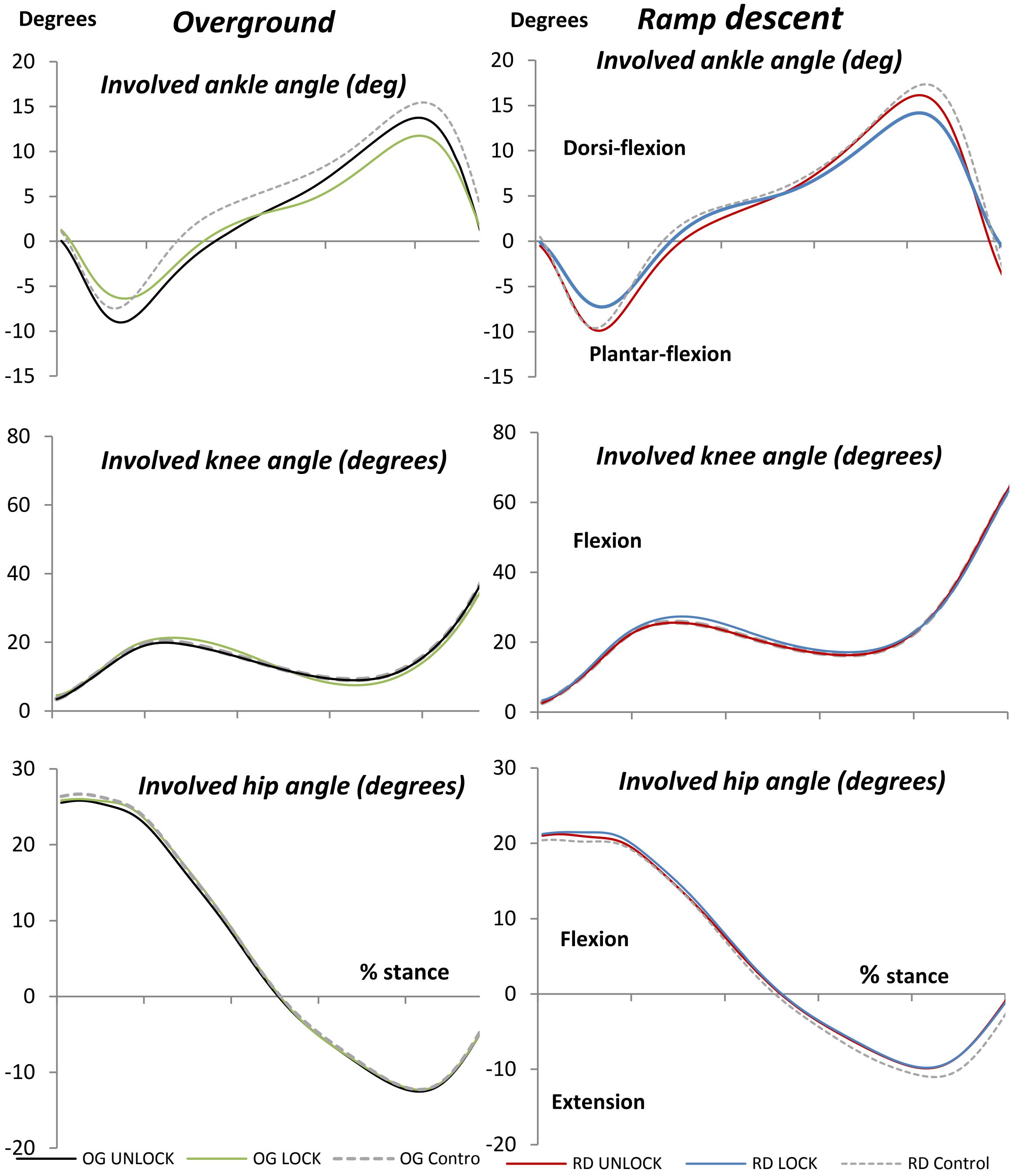

Figure 28. Involve limb (ankle, knee, hip) joints angular displacement (deg.) normalised to 100 points (stance phase), averaged across 20 participants. Positive angles are plantar-flexion and flexion for the knee and hip joints. (OG UNLOCK – (solid black line) overground non-restricted; OG LOCK – (solid green line) overground restricted; RD UNLOCK – (solid red line) ramp descent non-restricted; RD LOCK –(solid blue line) ramp descent restricted; OG Control - (dashed grey line) overground control data; RD Control (dashed grey line) ramp descent control data). NB for some of the figures the data for the different limbs appears not to be visible (included). This is because the angular displacement profile is very similar to another limb condition.

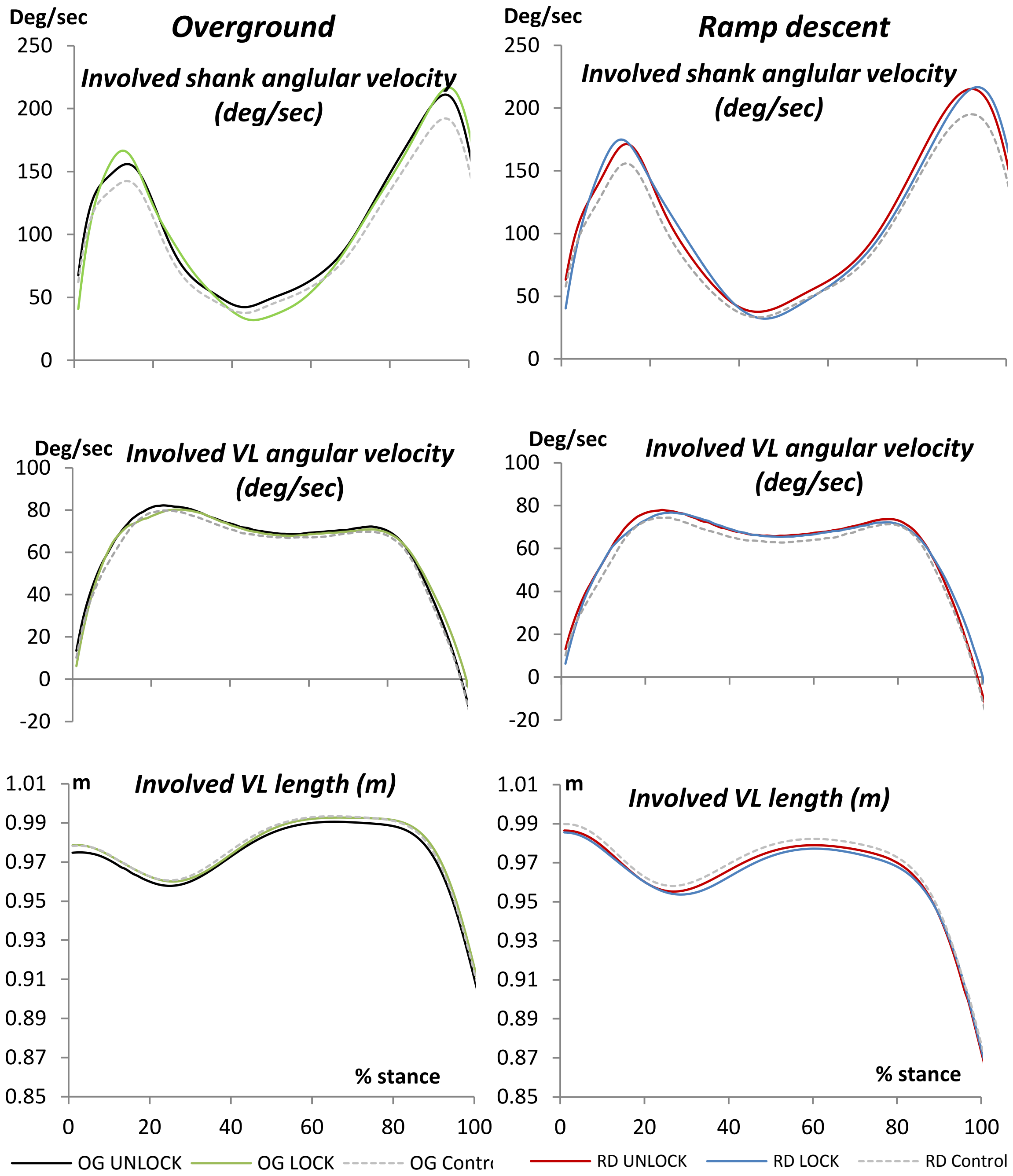

Figure 29. Mean Shank angular velocity, VL angular velocity, and VL length normalised to 100 points (stance phase), and ensemble averaged across 20 subjects. (OG UNLOCK –overground non-restricted; OG LOCK –overground restricted; RD UNLOCK –ramp descent non-restricted; RD LOCK –ramp descent restricted). OG LOCK – (solid green line) overground restricted; RD UNLOCK – (solid red line) ramp descent non-restricted; RD LOCK – (solid blue line) ramp descent restricted; OG Control – (dashed grey line) overground control; RD Control – (dashed grey line) ramp descent control. NB for some of the figures the data for the different limbs appears not to be visible (included). This is because the angular displacement or displacement profile is very similar to another limb condition.

Table 5. Group mean (±SD) involved (right) side: step length (m), stance time (m), knee loading response (deg.), CoM velocity throughout the stance (ms-1) and for the period of the single-limb-support (SLS) with AFO in non-restricted and restricted conditions in overground and ramp descent. Where differences are significant effect sizes Cohen's (d) are presented (in *italics*).

	Overground		**Ramp descent**		
	Non-restricted	*Restricted*	*Non-restricted*	*Restricted*	p value
Step length (m)	0.71 (0.07)	0.71 (0.06) *0.1*	0.66 (0.05)	0.65 (0.05) *0.2*	**Level<0.001** **cond. 0.01** Int. 0.57
Stance time (s)	0.713 (0.055)	0.718 (0.061) *0.1*	0.706 (0.055)	0.708 (0.065) *<0.1*	Level 0.18 cond. 0.17 Int. 0.53
CoM mean velocity throughout the stance (ms^{-1})	1.28 (0.14)	1.27 (0.15) *0.1*	1.20 (0.15)	1.19 (0.16) *0.1*	**Level 0.004** cond.0.20 Int. 0.31
CoM velocity throughout the SLS (ms^{-1})	1.25 (0.15)	1.24 (0.17) *0.1*	1.19 (0.16)	1.19 (0.17) *<0.1*	**Level 0.004** cond.0.28 Int. 0.23

4.3.2 Non-involved side (left limb)

The non-involved limb joints angular displacements during overground and ramp descent with restricted and non-restricted conditions are illustrated in Appendix 7. Appendix 8 illustrates the mean values of shank angular velocity. The mean (±SD) and statistical significance of step length, stance time, CoM velocity throughout the stance and for the period of the single-limb-support (SLS) on a non-involved (left) limb with AFO in non-restricted and restricted conditions in overground and ramp descent is presented in Table 6. There was no significant effect of ankle condition or interactions between the level of ambulation and ankle condition, so this will not be presented further in the results section unless stated otherwise.

Table 6. Group mean (±SD) Non-involved (left) side: step length (m), stance time (m), knee loading response (deg.), CoM velocity throughout the stance (ms-1) and for the period of the single-limb-support (SLS) with AFO in non-restricted and restricted conditions in overground and ramp descent.

	Overground		**Ramp descent**		p value
	Non-restricted	*Restricted*	*Non-restricted*	*Restricted*	
Step length (m)	0.70 (0.05)	0.69 (0.05)	0.65 (0.06)	0.64 (0.05)	**Level<0.001** cond. 0.08 Int. 0.22
Stance time (s)	0.730 (0.059)	0.739 (0.065)	0.718 (0.055)	0.722 (0.058)	**Level 0.04** cond. 0.10 Int. 0.31
Loading-response Knee flexion (deg.)	20.0 (6.6)	19.5 (6.4)	25.1 (6.9)	25.0 (6.6)	**Level<0.001** cond. 0.47 Int. 0.13
CoM mean velocity throughout the stance (ms^{-1})	1.27 (0.16)	1.26 (0.18)	1.20 (0.17)	1.20 (0.16)	**Level 0.006** cond.0.32 Int. 0.68
CoM velocity throughout the SLS (ms^{-1})	1.22 (0.17)	1.21 (0.18)	1.20 (0.17)	1.20 (0.16)	Level 0.64 cond.0.20 Int. 0.48

4.4 Discussion

The findings from the study suggest that the restricted ankle for overground and ramp descent has an influence on dynamic stability during the stance phase. Restricted ankle reduces the VL angular velocity at the beginning, but is unaffected at the end of SLS. The body motion changed according to gait mode and adapted lower-limb to the restricted ankle. The loading response knee flexion was increased in ramp descent compared to overground and increases to compensate restricted ankle. The spatial-temporal results also indicated that ramp descent and ankle restriction via AFO lead to a significant decrease in step length. Although, the ankle restriction did not affect the contralateral limb. The adaptations which caused by restricted ankle and/or ramp descent are discussed below and compared to the gait data that is reported in the literature.

The restriction of the ankle with AFO in both gait modes has a reduced range of motion plantar/dorsi-flexion (Figure 28). In ramp descent compared to overground gait the ankle has increased the range of dorsi-flexion and plantar-flexion. This also accords with a number of researchers' observations (Kuster et al. 1995; Lay et al. 2006; McIntosh et al. 2006) which reported that ankle dorsi-flexion increases to absorb increased gravitational potential energy during ramp descent. The function of the ankle together with knee joints in early stance contributes to control and absorption of the gravitational potential energy during ramp descent (Lay et al. 2006; McIntosh et al. 2006). Ramp descent compared to overground gait increased contribution predominantly at the knee and ancillary at the ankle into control and absorption. This is also supported in the number of studies (Donelan et al. 2002a; Lay et al. 2006; McIntosh et al. 2006; DeVita et al. 2007; Lay et al. 2007). Increased gravitational potential energy during ramp descent requires the contribution of ankle and knee joints increase to control and redirection pendulum transition. The function of knee and ankle was to redirect BW for subsequent pendulum transition to achieve efficiencies of a step-to-step transition. Hence, that efficiency depends on the functionality of knee and ankle joints. Pendulum transition to a subsequent limb requires redirection of the CoM velocity from one pendulum arc to the next.

The study results indicated the restricted ankle reduces VL angular velocity at the begin ($p < 0.01$) but no effect throughout or at the end of single-limb-support ($p < 0.11$ and effect size small $d \leq 0.1$). This was likely a result of the inability of the restricted ankle joint to plantar-flex within the function of propulsion at the begin of the stride prior to the beginning of single-limb-support (assessed stance phase on the force plate). Interestingly, ankle condition during the following SLS phase does not affect body motion (the VL angular velocity) was likely a result of adaptation at the knee joint as the shank angular velocity ($p<0.001$ and effect size $d \geq 0.3$ for both gait modes). Hence, individual joints of the lower-limb system adapt to ankle restriction to maintain body motion. The adaptation according to ankle condition seems to appear at the knee of the limb with the AFO. The knee loading response flexion was increased: firstly as a result of increased requirements to control BW motion forward/downward during ramp descent and secondly due to the inability of the ankle to plantar-flex in restricted condition. The increase of knee loading response during ramp descent with the assistance of gravity indicates that the knee provides a controlled strategy which was employed for a downward and forward transition. This finding is in agreement with Wall's et al. which suggested that knee loading response flexion increases with the increase of CoM vertical displacement as an impact force at the foot contact (Wall et al. 1981; Leroux et al. 2002; Hong et al. 2014). This seems to confirm the idea that gravity aids fall from the contralateral limb pendulum model, so the ankle and knee have to absorb potential gravitational energy within the control body motion downward/forward. To control body motion require flexion of the knee and plantar-flexion of the ankle to establish foot-flat. The increase of knee flexion during ramp descent is the result of the increased of the gravitational potential energy during ramp descent (Lay et al. 2006). The manipulation of functionality in one of the joints effect would lead to compensation at the remaining joints (Winter et al. 1990). Hence, participants would alter knee flexion according to ankle condition to maintain the whole body motion. The mechanism is more likely to be a repercussion of initial double support phase that participants encounter with increased knee flexion. This result corroborates with the findings of a great deal of the previous work in this field, so an increase of knee involvement with restricted ankle was presenting in TTs, which have increased knee loading response flexion with rigid ankle-foot prosthesis compared to able-bodied individuals (Vickers et al. 2008). The residual knee was being ‘thrown/pushed’ forwards to achieve foot-flat sooner.

Surprisingly, participants maintained a steady walking speed with both ankle modes the reason for this was likely to be because participants with restricted ankle alter their remaining joints kinetic and kinematic involvement to maintain an established self-selected walking speed. The findings of the current study do not support the previous research where ankle restriction led to reduced walking speed (Murray et al. 1984; Opara et al. 1985; Carlson et al. 1997; Romkes and Schweizer 2015). There are several possible explanations for this result. The calibrated volume did not allow recording full gait cycle so cadence was not assessed and the assessment of walking speed was done throughout the stance of a limb with the AFO. As the result walking speed was mainly determined by contralateral limb, restricted ankle would have only reduced the effect. In fact, the study participants were active males who had established walking speed, but reduced step length with restricted ankle could lead to an increase of cadence (steps per minute) (not determined in this study) within compensations in remaining joints. In addition, restricted ankle mode was not immobilised and had restricted ankle motion in the sagittal plane to around ±3-5 degree plantar/dorsi-flexion between participants.

The pendulum model provides an understanding of the mechanism of locomotion where CoM transfers over SLS and double support acts as a redirection from one arc to another. In the pendulum model, the ankle acts as the fulcrum. Initial double support begins with the collision and redirection of the arc forward and upward velocity. In a pendulum model during ramp descent compared to overground gait it had to increase control of BW transition over the support limb as BW transition effected by the gravitational potential energy. The limb supports BW and acts as a pendulum that conserves mechanical energy and without additional energy transfers BW over the support limb (Alexander

1991). To transfer the BW over the support foot with relatively minimal muscle involvement, the knee flexion of the support leg has to be minimal (Kuo 2007). Hence, the BW is supported passively during single-limb-support as a flexed leg would increase energy expenditure (Kuo et al. 2005). Certainly, the pendulum model has some drawbacks as a limb is not rigid so the arc of BW motion would be flawed. Also, that model could be applied only to the single-limb-support phase. However, using the knowledge from this drawback the motion of the pendulum could be predicted.

The VL angle at contralateral TO and IC were less during ramp descent compared to overground gait ($p < 0.01$), due to the shorter step length ($p < 0.01$). The reduction of step length could state the changes in CoM position in relation to support feet positions. The restricted ankle also reduces the VL angle on the contralateral TO ($p < 0.001$ and effect size ($d=0.3$) small to medium) for both gait modes. This was likely a result of a reduction in the stride length (however, it was not tested in this study). Because at the beginning of the stride (lateral foot ‘push off’ prior IC on the force plate), the restricted ankle could not provide necessary planter-flexion for the foot swing and body propulsion. Thus, individuals with restricted ankle have reduced step length, which seems to be a result of the restricted ankle ‘push off’ phase. The study findings would suggest, the ‘push off’ at lateral limb is critical for sufficient step length and comparable between overground and ramp descent. Ramp descent compared to overground and restricted compared to non-restricted ankle condition led to the step length reduction. The research of Pijnappels et al. was experimentally measured and indicated that reduced propulsion would affect stride length (Pijnappels et al. 2005). Hence, the use of the VL angles at contralateral TO and IC could expand the view of pendulum transition over the support limb and provide additional information to clarify gait pathology.

The use of the VL length provides additional knowledge and estimates the combined function (flexion/extension) of the hip and knee in the lower-limb system. The support limb (VL length) during ramp descent compared to overground gait was lengthened in early stance, but shortened to lower down the BW ramp descent mid and late stance (Figure 29). Down the ramp, the VL length was lengthened in early stance to maximise limb length for upcoming absorption. Hence, participants during ramp descent seem like changing control strategies in the swing phase, which is observed during level walking and pre-plan lowering the BW. The lengthened VL parameter would increase the range for absorption in the limb. The shortening of VL length during mid-stance could indicate, that ramp descent compared to the overground gait has increased absorption, which is followed by the CoM lowering to reduce 'fall' on a contralateral limb in late stance. However, manipulation with ankle restriction has negligible effect on the VL length (Figure 29).

The importance of attaining foot-flat sooner for overground gait and ramp decent is highlighted below. Time to attain a foot-flat was unaffected by gait mode ($p = 0.13$) which likely explains why the knee involvement was increased in ramp descent. The attained foot-flat provides support for BW transition in conjunction with the vertical CoM displacement that occurred as BW falls from the contralateral limb. To reduce the impact, participants' likely reduces a step length ($p<0.001$) during ramp descent compared to level walking. Similarly, Redfern and DiPasquale (1997) proposed that the step length during ramp descent lessened the load on the lead limb (Redfern and DiPasquale 1997). Therefore, this adaptation of the foot placement was performed to ensure the load on lower-limb joints is within comfortable and safe boundaries. The data reported here appear to support the assumption that the shorter step length on inclined surfaces enhances anterior-posterior stability (Silverman et al. 2012) and ensures dynamic stability (Kawamura et al. 1991; Sun et al. 1996; McIntosh et al. 2006). Restricted ankle delayed time to attain foot-flat ($p < 0.001$ and effect size large $d \geq 0.7$) was likely to lead to an increase in knee flexion loading response ($p < 0.001$). The attainment of the foot-flat sooner is critical to provide a stable base of support for the BW transitions ('falls') from the contralateral limb to the lead limb that is wearing the AFO. Participants' with restricted ankle could not plantar-flex in order to attain a foot-flat after IC so as a result to attain of foot-flat would be delayed. Hence, participants with restricted ankle are unable to articulate to achieve foot-flat sooner, and as a consequence the song 'pulls' forward during SLS ($p < 0.001$ and the effect size is large d=0.6 in overground but small d=0.1 in ramp descent) and increased knee loading response by ~5-6% ($p < 0.001$). This finding supports previous research, TTs have increased knee involvement (rigid ankle-foot prosthesis) compared to healthy individuals as the residual knee was being 'thrown/pushed' forwards in order to achieve foot-flat sooner (McIntosh et al. 2006; Vrieling et al. 2008). Another study has demonstrated that when the TTs are walking down slopes, the attainment of the foot-flat depend on articulation in ankle-foot devices (Vickers et al. 2008). Therefore, to attain foot-flat the shank pulling forward with the increase of knee loading response which possibly could reduce dynamic stability. The increase of knee loading response requires eccentric strength in knee extenders that could possibly reduce knee stability. This could be critical for TTs due to reduced muscle volume and strength in the residuum (Perry et al. 1997; Isakov et al. 2000; Vickers et al. 2008). Interestingly, powered ankle-foot prosthetic device Proprio-Foot from Ossur has articulation during the swing phase. The ankle-foot device 'plantar-flex' during ramp descent to ensure appropriate accommodation on inclined surfaces (Versluys et al. 2008; Eilenberg et al. 2010). Hence, during ramp descent, the stance phase of the lead/front limb was loaded within safe and comfortable boundaries. These findings suggest that the restricted ankle condition in both gait modes, if the foot was not able to plantar-flex after IC could affect dynamic stability in early stance phase.

4.5 Conclusions

The present study identified the significance of ankle motion during the stance phase overground compared to ramp descent and partly confirms the study hypothesis. The restricted ankle affects the shank angular velocity, but increased knee loading response was smoothing body weight transfer (VL angular velocity) relative to the support foot ankle at single-limb-support phase. Therefore, increased knee loading response is evidence of adaptations which is a compensatory mechanism to ankle restriction in both gait modes. To attain foot-flat quicker require for balance control, so participants' with the restricted ankle have reduced step length. In view of all that has been mentioned so far, one may suppose that the restriction of the ankle affects dynamic stability in both gait modes. It is clear that the results suggest no fundamental change in gait between overground and ramp descent with 5 degrees of inclination. Moreover, this study has highlighted that gait performance assessment together or instead of; conventional kinematic variables can be employed the VL variable that could provide simplified insights to the body motion relative to the support limb behaviour on the level and inclined surfaces.

CHAPTER FIVE - JOINT KINETIC ADAPTATIONS WHEN WALKING DOWN THE RAMP: EFFECTS OF UNILATERAL ANKLE BRACING ON ABLE-BODIED INDIVIDUALS

5.1 Introduction

The stance phase of overground gait can be described as involving three sequential functional rockers (Perry and Davids 1992). The first is associated with the foot (toe region) being lowered to the ground following heel contact ('heel' as a rocker) and body weight (BW) being accepted onto the limb. The second describes the period of single-limb-support when the Centre-of-Mass (CoM) progresses forwards ('rolls') over the limb ('ankle' as a rocker) while the contralateral limb is swung forward. The third is associated with the transfer of BW off the limb ('fore-foot' as a rocker) onto the contralateral limb. The three-rocker model essentially describes an inverted pendulum (IP) (Alexander 1995; Kuo 2002) with the foot-ankle complex acting as the fulcrum. The first and third rocker phases represent periods where BW is transferred onto and off the limb, with predominantly negative (eccentric) joint work done during the first rocker and positive (concentric) joint work done during third rocker (Donelan et al. 2002b). The progression of the CoM over the limb during the second rocker period occurs predominantly passively, i.e. without any significant joint work (Winter 1983) and involves pendulum motion over the foot.

The gait involved in walking down slopes can also be described using the three-rocker model (IP). However, because there is a requirement also to lower the CoM as it progresses forward, the lower-limb joints work contributions are different from that in overground gait (Lay et al. 2006). During the first rocker period in comparison to overground gait, more negative work occurs at the ankle following the instant of ground contact as the fore-foot has to be lowered further to achieve a 'foot-flat' position on the ramp, and more negative work is required at the knee during weight acceptance to control the increased lowering of the CoM (Lay et al. 2007). The ankle also does more negative work during the second rocker, in comparison to that in overground gait, in order to control the rate of forward shank rotation as the CoM progresses over the limb while being lowered down the ramp (McIntosh et al. 2006; Lay et al. 2007). During third rocker gravity will assist the transfer of BW onto the contralateral limb (Lay et al. 2007) so the ankle push-off requirement is reduced, and hence less positive ankle work is done compared to overground gait (Franz et al. 2012).

The paragraph above highlights the importance of ankle motion to the stance phase of gait. Hence, ankle motion will likely have a significant impact on all of the three rocker stages of gait. As the ankle is required to exert more control when walking down slopes, ankle functionality is likely to have a greater impact on downslope gait than on overground gait. This would explain why ankle function is important during descending slopes.

In order to better understand the compensatory joint kinetics used by those with unilateral ankle amputations when walking down slopes, the present study determined how the unilateral restriction of ankle motion in healthy young adults affected joint kinetics for downslope gait in comparison to overground gait. Ankle motion was manipulated by use of an ankle-foot orthosis which restricted ankle motion in the sagittal plane to around ± 3-5 degrees in plantar/dorsi-flexion which was estimated between all able bodied participants. It was hypothesised that ankle bracing in overground gait would have little effect on the ankle work done during the 1st and 2nd rocker periods, but would reduce the amount of positive ankle work done during the 3rd rocker period. It was also hypothesised that bracing of the ankle in downslope gait would prevent the ankle doing the anticipated increased eccentric work during the 1st and 2nd rocker periods, and in compensation, the knee would do more negative work during these periods. However, due to gravity assisting the transfer of body weight (BW) onto the contralateral limb, ankle bracing would not affect the positive ankle work done during this period.

5.2 Methods

5.2.1 Participants

Twenty physically active, males (mean (SD) age 27.5 (8.0) years, mass 84.5 (11.5) kg, height 1.79 (0.06) m), participated, each indicating they had no gait impairments, musculoskeletal disorders or history of major injury to the lower-limbs. Full details are presented in chapter 3.4.

5.2.2 Specific equipment, procedure, data acquisition and processing

To manipulate ankle articulation, the custom made ankle-foot orthosis (AFO) device(Figure 22) was utilised. Details of the AFO and walking procedures are presented in Chapter 3.11

To examine ramp descent, a custom made modular ramp 2.8 metres long, with an inclination of 5 degrees and 1.0-metre long level ground landing was used (Figure 19). Full details are provided in chapter 3.7.6. Prior data collection and when the solid incline block was removed or installed on the force plate (FP), to ensure correct data the FP was zeroed in Vicon Nexus 1.8 software (Vicon, Oxford, UK) and amplifier was reset. Ten Vicon MX cameras used to capture motion during the overground and ramp descent walking trials (Chapter 3.7.2). The cameras were positioned in a circle surrounding the FP, which is located the centre of the laboratory, to minimise reconstruction error within a calibrated volume. The calibrated volume had dimensions of approximately 3 m (length) x 2 m (width) x 2.5 m (height) (Figure 19). The volume was calibrated prior each data collection using the Marker Wand (390 mm) where Clinical L-frame (Figure 15) used to set up the orientation of the cameras in 3D perspective within XYZ coordinate origin as described in chapter 3.7.2. The volume calibration and the coordinate origin set up were performed without the ramp.

Details of all laboratory equipment used, how kinetic and kinematic data were recorded with the full protocol procedure are presented in chapter three. Participants were identified by passive retro-reflective markers according to six degrees of freedom (6 DoF) model as described in chapter three.

Participants were introduced to the laboratory and prior to recording kinematic and kinetic data, allowed to familiarise themselves with the walking tasks and ‘practice’ typically two trials, according to the data collection protocol (Chapter 3.11). Each participant completed two blocks of repeated gait trials, one involving walking down the ramp and the other along the level ground (i.e. laboratory floor without a ramp). Block order was counterbalanced across participants. Each block included two ankle conditions, restricted and non-restricted, the order of which was counterbalanced across participants. Each participant completed six successful trials (landing precisely on of the force platform without gait alterations) with the involved and non-involved limbs in restricted/non-restricted modes in counterbalance order among the participants. All conditions were repeated six times; hence the total number of trials completed was 48: 6 (repetitions) x2 ankle conditions (non-restricted and restricted), x2 limbs (involved with AFO, non-involved), x2 gradients (overground and downslope). Participants were instructed, which limb they should initiate gait with prior to each trial and to walk at speed they would normally walk, i.e. at their freely chosen speed as described in chapter three.

Sagittal plane joint (muscle) moments were determined using standard inverse dynamics (Winter 2009). Thus moment of the ankle with AFO was determined for the whole system, i.e. orthotic plus biological joint. The associated joint powers were determined as the product of the net joint moment and angular velocity at the assessed joint (Equation 3):

$$P = M \times \omega_s \qquad (1)$$

Where M is the sagittal joint moment (Nm) acting at the proximal end of a segment and ω_s is joint angular velocity (rad.s^{-1}) between the two segments intersecting at the joint.

Negative and positive joint work were determined as the integrals of the negative and positive sections respectively of the joint power curve.

The detailed description of data recording, processing and filtering provided in chapter three.

Initial contact (IC) and ipsilateral toe-off (TO), were determined from force platform. Contralateral limb IC (IC_{con}) and TO (TO_{con}) were determined from kinematic data. Description and elucidation for those gait events are presented in Chapter 3.13.

5.2.3 Data analysis

The following parameters were determined for each trial. Lower-limb joint moments and powers were normalised to body weight and height. Lower-limb joint positive and negative work were integrals respectively, for DS1, SLS, DS2 phases. Positive and negative joints work was the positive and negative joint power integrals respectively, for DS1, SLS, DS2 phases. Total limb scalar joint work was the sum of the positive and negative work for all joints (ankle, knee, and hip). Mean values for each participant were determined for each gait mode and ankle condition.

5.2.4 Statistics

Data were analysed using repeated measures analysis of variance (ANOVA) with ankle condition (non-restricted, restricted) and surface condition (overground, downslope) as repeated factors. The analyses of limb work (positive, negative and total) throughout the 3 phases (DS1, SLS, DS2) as covariates were repeated as analyses of covariance (ANCOVA). Effect size differences (low $d < 0.3$, moderate $0.3 < d < 0.5$ and high $d < 0.5$) were calculated as Cohen's (Cohen 1988). Statistical analyses were performed in Statistica (v6, StatSoft, Inc., Tulsa, OK, USA). A post hoc analysis was undertaken using Duncan tests. The level of significance was set $p < 0.05$.

5.3 Results

5.3.1 Involved limb: DS1: double support phase during weight acceptance

Ankle negative work reduced with ankle restriction ($p = 0.01$) and was greater for downslope compared to overground gait ($p < 0.001$); there was no interaction between terms ($p = 0.54$). The amount of knee negative work, hip positive and total limb negative joint work was unaffected by ankle restriction ($p \geq 0.43$), however, greater ankle, knee, and total limb negative joint work and less hip positive work was done for downslope compared to overground gait ($p < 0.001$); there were no interactions between terms ($p \geq 0.3$). Total limb scalar joint work was unaffected by ankle restriction ($p = 0.46$) but was reduced for downslope compared to overground gait ($p < 0.01$); there were no interactions between terms ($p = 0.47$).

5.3.2 Involved limb: SLS: single-limb-support phase

Ankle negative work reduced with ankle restriction (by ~14-15%, $p < 0.001$) and was greater for downslope compared to overground gait ($p < 0.001$); there was also an interaction between terms ($p = 0.016$). Ankle restriction leads to reduced ankle negative work during overground and downslope gait ($p < 0.001$). Knee positive work increased with ankle restriction (by ~20-22%, $p < 0.001$) but was unaffected by the surface condition ($p = 0.18$), and there was no interaction between terms ($p = 0.97$). Hip negative work was unaffected by ankle restriction ($p = 0.76$) or by the surface condition ($p = 0.41$); but there was a significant interaction between terms ($p = 0.05$). However, post hoc analysis indicated no significant differences between surface or ankle conditions ($p = 0.36$). Total limb negative joint work was unaffected by ankle restriction ($p < 0.03$) but was greater for downslope compared to overground ($p < 0.001$); there was also an interaction between terms (trend only, $p = 0.06$). Ankle restriction leads to greater total limb negative joint work during overground gait ($p = 0.02$) but had no effect in downslope gait ($p = 0.64$). Total limb scalar joint work was unaffected by ankle restriction ($p = 0.17$) but was reduced for downslope compared to overground gait ($p < 0.001$); and there was no interaction between terms ($p = 0.09$).

5.3.3 Involved limb: DS2: double support phase during unweighting

Ankle positive work reduced with ankle restriction (by ~17-19%, $p < 0.001$) and was decreased for downslope compared to overground gait ($p < 0.001$): there was no interaction between terms ($p = 0.32$). Knee negative work reduced with ankle restriction (by ~4-10%, $p < 0.01$) and was increased for downslope compared to overground gait ($p < 0.01$). There was also a significant interaction between terms ($p = 0.04$). Ankle restriction leads to reduced knee negative work during overground and downslope gait ($p < 0.03$). Hip positive work was unaffected by ankle restriction ($p = 0.98$) but was reduced for downslope compared to overground gait ($p < 0.001$); there was no interaction between terms ($p = 0.45$). Total limb positive joint work reduced with ankle restriction (by ~12-13%, $p < 0.001$) and was reduced for downslope compared to overground gait ($p < 0.001$); there was no interaction between terms ($p = 0.37$). Total limb scalar joint work reduced with ankle restriction ($p < 0.01$) and was reduced for downslope compared to overground gait ($p < 0.01$); there was no interaction between terms ($p = 0.93$).

Table 7. Group mean (±SD) involved (right) limb: ankle, knee and hip (positive and negative) work for initial double support (DS1), single-limb-support SLS) and terminal double-limb-support (DS2) for non-restricted and restricted ankle conditions in overground and downslope gait. Where differences between ankle conditions are significant effect sizes Cohen's (d) are presented (in *italics*).

	Overground		***Downslope***		*p value*
	Non-restricted	*Restricted*	*Non-restricted*	*Restricted*	
Ankle Work (J. kg^{-1}) DS1	-0.028 (0.019)	-0.021 (0.013) *0.4*	-0.041 (0.031)	-0.036 (0.025) *0.2*	**level<0.001** **cond. 0.01** Int. 0.54
SLS	-0.246 (0.048)	-0.211 (0.046) *0.7*	-0.312 (0.053)	-0.262 (0.048) *0.6*	**level<0.001** **cond.<0.001** **Int. 0.02**
DS2	0.199 (0.067)	0.164 (0.060) *0.5*	0.146 (0.056)	0.110 (0.05) *0.5*	**level<0.001** **cond.<0.001** Int. 0.32
Knee Work (J.kg^{-1}) DS1	-0.033 (0.033)	-0.036 (0.030) *0.1*	-0.121 (0.064)	-0.117 (0.055) *0.1*	**level<0.001** cond.0.90 Int. 0.30
SLS	0.052 (0.037)	0.066 (0.035) *0.4*	0.059 (0.037)	0.074 (0.036) *0.4*	level 0.18 **cond.<0.001** Int. 0.97
DS2	-0.162 (0.055)	-0.146 (0.049) *0.3*	-0.184 (0.064)	-0.177 (0.064) *0.1*	**level <0.01** **cond.<0.01** **Int. 0.04**
Hip Work (J.kg^{-1}) DS1	0.062 (0.023)	0.064 (0.021) *0.1*	0.030 (0.014)	0.031 (0.012) *0.1*	**level<0.001** cond.0.48 Int. 0.88
SLS	-0.161 (0.034)	-0.125 (0.036) *0.2*	-0.075 (0.051)	-0.101 (0.079) *0.4*	level 0.41 cond.0.76 **Int. 0.05**
DS2	0.085 (0.020)	0.084 (0.015) *0.1*	0.065 (0.018)	0.066 (0.018) *0.1*	**level<0.001** cond.0.98 Int. 0.45

5.3.4 Involved limb: limb rotational work

Limb negative rotational work has an interaction between terms: gait phase and gait mode ($p < 0.001$). Downslope compared to overground gait leads to increased limb negative rotational work during DS1 and SLS phases, but had no effect during the DS2 phase. There was also a significant interaction between terms: gait phase, gait mode and ankle condition ($p = 0.04$). Ankle restriction leads to reduced limb negative rotational work during the SLS phase in overground gait, but had no effect during downslope and had no effect during DS1 and DS2 phases. Downslope compared to overground gait lead to greater limb negative rotational work during DS1 and SLS phases, but had no effect during the DS2 phase. Limb positive rotational work has an interaction between terms: gait phase and gait mode ($p < 0.001$). Downslope compared to overground gait leads to reduced limb positive rotational work during the DS2 phase and had no effect during DS1 and SLS phases. Limb positive rotational work also has an interaction between terms: gait phase and ankle condition ($p < 0.001$). Ankle restriction leads to reduced limb positive rotational work during the DS2 phase, but had no effect during DS1 and SLS phases. Limb total rotational work has an interaction between terms: gait phase and gait mode ($p < 0.001$). Downslope compared to overground gait leads to increased limb total rotational work during DS1 and SLS phases, but had reduced during the DS2 phase. There was also a significant interaction between terms: gait phase and ankle conditions ($p = 0.04$). Ankle restriction leads to reduced limb total rotational work during the DS2 phase, but had no effect during DS1 and SLS phases.

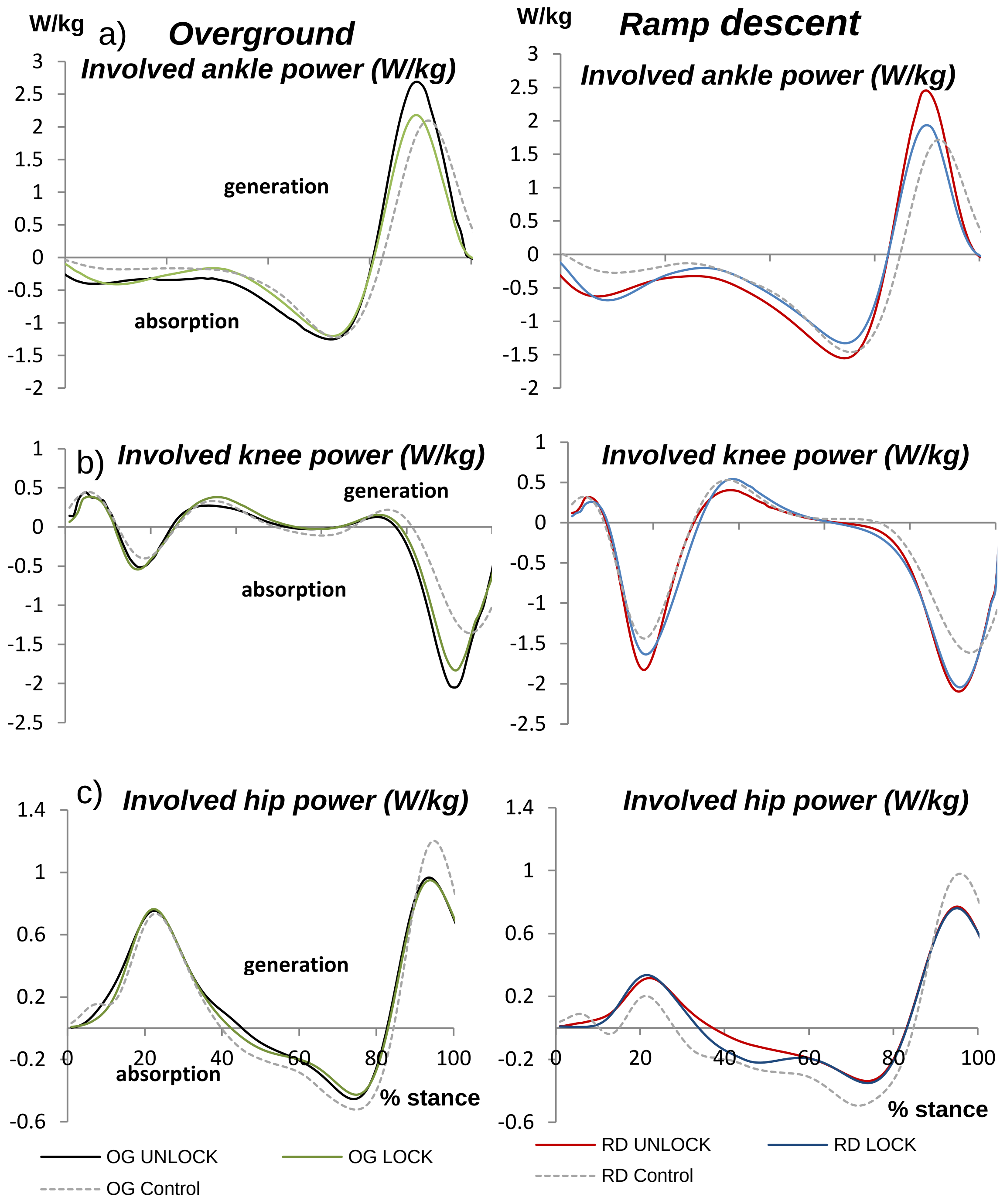

Figure 30. Mean of involved (right) limb: a/ ankle joint power (W/kg); b/ knee joint power (W/kg); c/ hip joint power(W/kg) normalised to 100 points (stance phase), and ensemble averaged across 20 subjects. (OG UNLOCK – (solid black line) overground non-restricted; OG LOCK – (solid green line) overground restricted; RD UNLOCK – (solid red line) ramp descent non-restricted; RD LOCK –(solid blue line) ramp descent restricted; OG Control - (dashed grey line) overground control data; RD Control (dashed grey line) ramp descent control data).

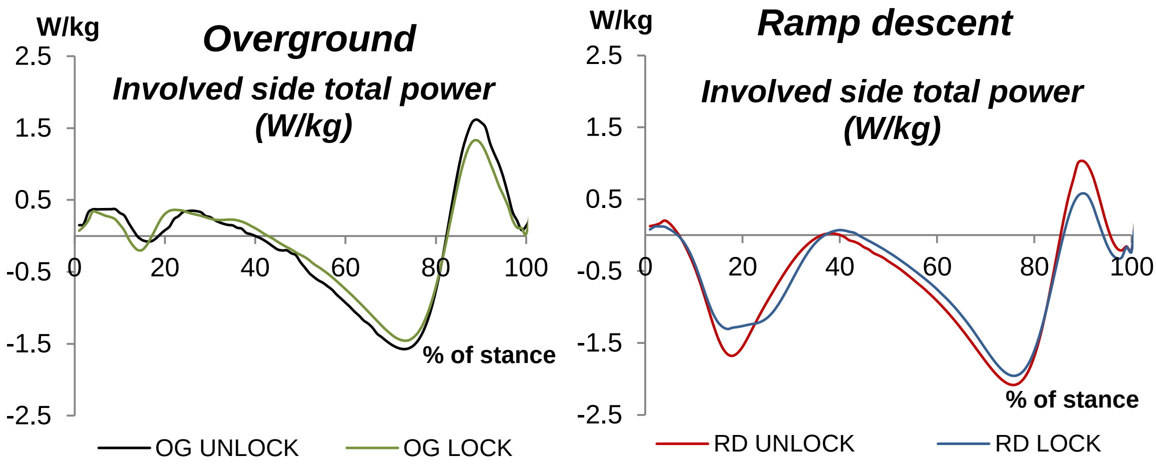

Figure 31 Mean of involved (right) limb total rotational power (W/kg) normalised to 100 points (stance phase), and ensemble averaged across 20 subjects. (OG UNLOCK – (solid black line) overground non-restricted; OG LOCK – (solid green line) overground restricted; RD UNLOCK – (solid red line) ramp descent non-restricted; RD LOCK –(solid blue line) ramp descent restricted).

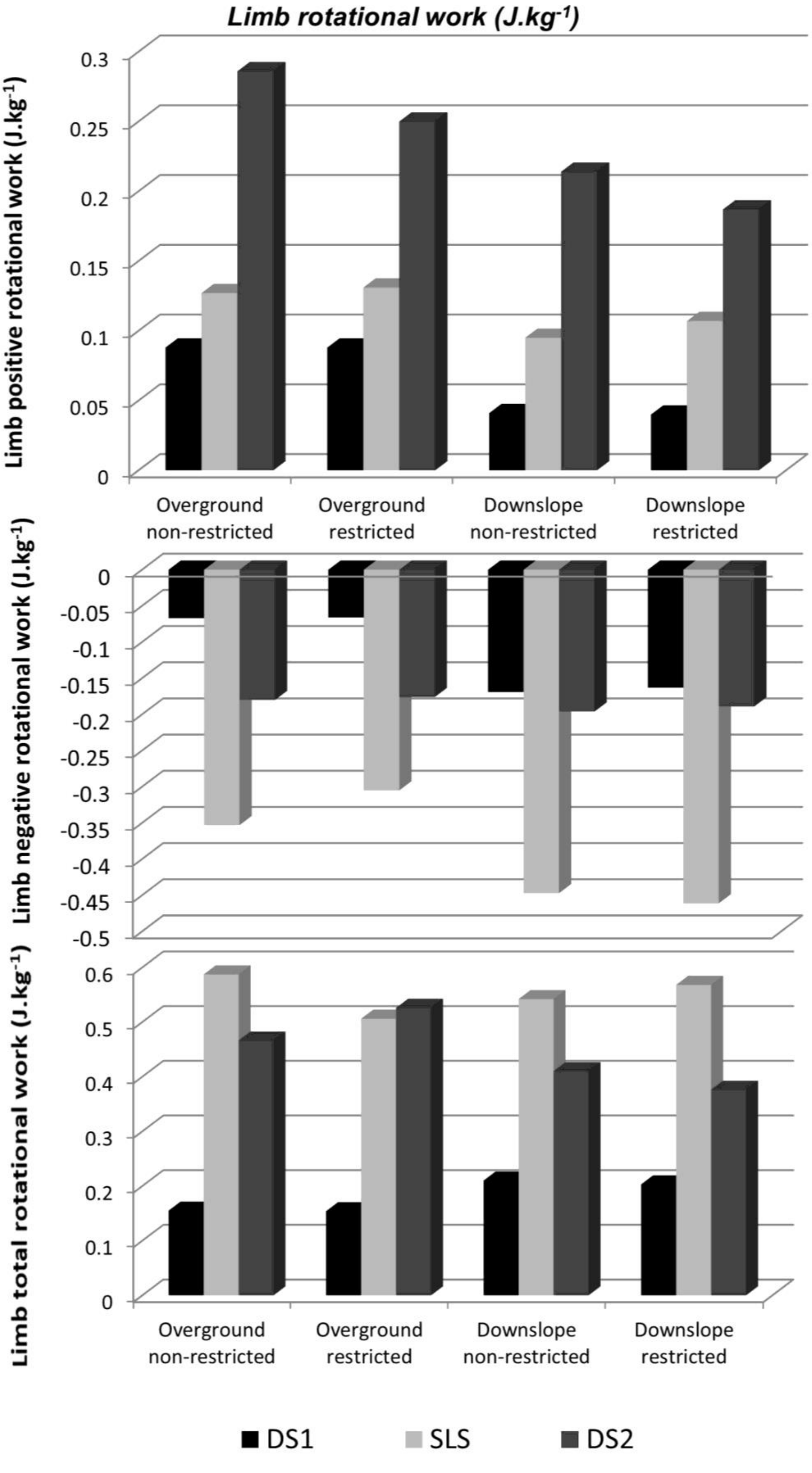

Figure 32. Mean: Limb positive rotational work (top left) (J.kg^{-1}); Limb negative rotational work (bottom left) (J.kg^{-1}); Limb total rotational work (right) (J.kg^{-1}); (during DS1, SLS, DS2 rockers), and ensemble averaged across 20 subjects.

5.3.4 Non-involved side (left limb)

The non-involved limb joint powers during overground and ramp descent with restricted and non-restricted conditions are illustrated in Appendix 9. The mean (±SD) and statistical significance of the ankle, knee, and hip for the period of DS1, SLS and DS2 on a non-involved (left) limb with AFO in non-restricted and restricted conditions in overground and ramp descent is presented in Table 8. There was no significant effect of interactions between the level of ambulation and ankle condition, so this will not be presented further in the results section unless stated otherwise. Hip positive work for the period of DS1 reduced with ankle restriction ($p = 0.01$) and reduced for the period of DS2 compared to non-restricted ankle condition ($p = 0.04$).

Table 8. Group mean (±SD) non-involved (right) limb: ankle, knee and hip (positive and negative) work for initial double support (DS1), single-limb-support SLS) and terminal double-limb-support (DS2) for non-restricted and restricted ankle conditions in overground and downslope gait. Where differences between ankle conditions are significant effect sizes Cohen's (d) are presented (in *italics*).

	Overground		***Downslope***		*p value*
	Non-restricted	*Restricted*	*Non-restricted*	*Restricted*	
Ankle Work (J. kg^{-1}) DS1	-0.018 (0.012)	-0.016 (0.011) *0.2*	-0.031 (0.021)	-0.030 (0.018) *0.1*	**level<0.001** cond. 0.14 Int. 0.48
SLS	-0.233 (0.039)	-0.223 (0.041) *0.1*	-0.274 (0.043)	-0.251 (0.037) *0.2*	**level<0.001** cond.0.15 Int. 0.11
DS2	0.168 (0.057)	0.164 (0.055) *0.1*	0.120 (0.053)	0.130 (0.060) *0.2*	**level<0.001** cond.0.59 Int. 0.15
Knee Work (J.kg^{-1}) DS1	-0.030 (0.034)	-0.027 (0.032) *0.1*	-0.114 (0.081)	-0.113 (0.076) *0.1*	**level<0.001** cond.0.31 Int. 0.76
SLS	0.043 (0.029)	0.056 (0.031) *0.1*	0.060 (0.041)	0.065 (0.039) *0.2*	level 0.11 cond.0.12 Int. 0.34
DS2	-0.129 (0.058)	-0.126 (0.057) *0.1*	-0.177 (0.078)	-0.172 (0.082) *0.1*	**level <0.001** cond.0.48 Int. 0.92
Hip Work (J.kg^{-1}) DS1	0.055 (0.021)	0.051 (0.021) *0.2*	0.024 (0.021)	0.020 (0.018) *0.2*	**level<0.001** **cond.0.01** Int. 0.97
SLS	-0.153 (0.039)	-0.138 (0.040) *<<0.1*	-0.109 (0.049)	-0.121 (0.058) *0.1*	level 0.21 cond.0.36 Int. 0.09
DS2	0.103 (0.024)	0.101 (0.025) *0.1*	0.085 (0.035)	0.078 (0.032) *0.2*	**level<0.001** **cond.0.04** Int. 0.36

5.4 Discussion

The present study determined that the use of unilateral ankle restriction was leading to compensations in the remaining joints and those compensations are distinct between downslope and overground gait. Restriction of the ankle affected its contribution by the reduction of negative ankle work for a period of 1^{st}, and 2^{nd} rocker phases and also reduction of positive ankle work done for a period of 3^{rd} rocker in both gait conditions, which only partially supported the hypothesis. The study findings indicated that downslope compared to overground gait has increased the negative knee work, but reduced positive work at the hip. The increased negative work at the knee downslope requires a good knee muscle extensors (eccentric strength). This explains, why slope descent compensations could have a detrimental effect on TTs. Restriction of the ankle led to the compensations at the involved knee joint where concentric work increased during 2^{nd} rocker, but reduced eccentric work during 3^{rd} rocker for both levels of ambulation, also only partially supported the hypothesis. The examination of limb total negative and positive rotational work reflected general limb performance under the changed levels of ambulation and ankle conditions. Restriction of the ankle reduces the positive work on the contralateral hip for a period of 1^{st} and 3^{rd} rockers for both levels of ambulation. The results highlight that ankle restriction had a non-significant effect between overground and slope descent with 5 degrees of inclination.

The study results indicated that downslope compared to overground gait has increased limb negative rotational work during 1^{st} rocker, where only the ankle and knee contribution in such negative joint work. There negative joint work increased primarily at the knee by ~71% and secondary at the ankle by ~36 %. To achieve controlled BW transition with increased potential gravitational energy during slope descent requires increased at the ankle and knee joint contribution. These results are consistent with those of other studies and suggest that downslope compared to overground gait requires an increase of body downward motion control in early stance, where to control body motion predominantly used the knee and ancillary the ankle joint (Lay et al. 2006;

McIntosh et al. 2006; DeVita et al. 2007; Lay et al. 2007). The increase of negative knee work done during the 1st rocker is likely to correspond with the increase of loading response knee flexion (Chapter Four). Hence, the knee primary contributors to acceptance of BW which was driven by CoM vertical displacement. In downslope gait, increased CoM vertical displacement supplies additional energy in BW transition (Garcia et al. 1998) where the vertical displacement depends on approached slope gradient. Because the knee extensor eccentrically controls BW acceptance, slope descent could be problematic for TTs due to the partly amputated or weakness of these muscles (Winter and Sienko 1988). To absorb the collision and provide controlled and smoothed this BW transition over the support limb after initial contact also requires an increase of ankle eccentric work contribution. Those results matched those observed in earlier studies (Wall et al. 1981; Leroux et al. 2002; Hong et al. 2014). These effects of the restricted ankle in able-bodied are similar to the effects of rigid 'ankle' in unilateral TTs. Both have malfunctioned in ankle motion. Contrary to expectations, as TTs have an increase of work at the contralateral hip in early stance (Silverman et al. 2008). The study results indicated that restricted ankle reduces the contralateral hip work for a period of 3rd rocker ($p = 0.01$) (stance phase of a contralateral limb), but there is a low effect size ($d \leq 0.2$). This case reveals the need for further investigation in able-bodied individuals with the restricted ankle.

Participants in downslope gait have increased limb negative rotational work in 2nd rockers to control BW motion within the gravity assistance which was likely to correspond to reduced walking speed (stance) (Chapter Four). This result corroborates the ideas of Kuo and Donelan (2010), who suggested that the negative work of the lower-limb is dependent on the amount of CoM vertical displacement in the arc of the pendulum model (Kuo and Donelan 2010). The study results indicated that the knee positive and hip negative work was not affected by gait mode ($p \geq 0.18$) for a period of 2nd rocker. Hence, the ankle joint provides primary control of BW transition in both gait modes. If in overground gait ankle joint acts as the fulcrum with the objective to support the CoM passively without active control and with minimal or no energy expenditure

(Alexander 1995; Kuo 2002) but slope descent would require control from the ankle. Nevertheless, manipulation with the ankle motion by restriction led to the knee positive work increase by ~19-21% ($p < 0.001$ and effect size medium $d=0.4$) which was likely a counter to the knee loading response flexion which was increased in both gait modes (Chapter Four). As a consequence, participants tended to return the limb to the optimal/efficient length, which increases the knee joint concentric work which was likely to increase the energy transfer as a result of interfering with transition efficiency. This finding suggested that the restriction of the ankle articulation likely has an effect on energy expenditure as a result of interruption of the normal gait cycle, which collaborates with other studies that examined overground gait (Neptune et al. 2001; Donelan et al. 2002b). This interruption in BW transition by the ankle restriction would also lead to compensatory mechanisms, which is, however, aimed to minimise increased energy expenditure (Inman et al. 1981).

The period of the 3rd rocker, negative knee work has increased during downslope gait ($p < 0.01$). There knee eccentric work contributes to lowering the CoM for the subsequent step. This result matches those observed in the research of Franz et al. (Franz et al. 2012). Controlled lowering of the BW could be challenging for TTs due to the residuum weakness and/or deficiency of muscle at the knee joint (Winter and Sienko 1988; Perry et al. 1997; Isakov et al. 2000; Vickers et al. 2008). The BW downward/forward transition aided by increased potential gravitational energy for downslope gait (Chapman 2008) with transfer onto the contralateral limb and led to a reduction of push-off requirements from the ankle during 3rd rocker ($p < 0.001$). Although, gait downslope (5 degrees) still requires push-off from the ankle to transfer BW for the subsequent step and/or prepare the limb for a swing. The results confirmed what previous studies had shown; power generation requirements decreased during downslope gait as gravity assists BW transition (Lay et al. 2007; Franz et al. 2012). The study results support that, the limb positive and limb total rotational work was reduced for downslope compared to overground gait ($p < 0.01$). The restriction of the ankle reduced the amount of late stance ankle power, which is known to be related to forward propulsion (Lehmann 1993).

Surprisingly, the restricted ankle still has reduced knee negative work in the 3rd rocker as compensation for both gait modes. Although, this compensation could be a result of different causes. A possible explanation for this, in overground gait, might be due to the inability of the ankle to plantar-flex in order to propel BW forward in the restricted condition. However, in downslope gait, the result may be explained by the fact that restricted ankle is unable to dorsi-flex on an inclined surface, so the knee reduces compensation. Nevertheless, the compensation to the restricted ankle was done on the involved limb knee joint, which was increased when slope descent was approached. Surprisingly, the contralateral hip for a period of 1st rocker ($p = 0.01$) (stance phase on a contralateral limb) reduces positive work to the restricted ankle which is assisting the body to vault over the stance limb. These results differ from the study with TTs participants where contralateral hip has increased work in late stance (Silverman et al. 2008). However, hip positive work has low effect size ($d \leq 0.2$) but there is a need for further investigation in able-bodied individuals with the restricted ankle. Therefore, the ankle motion has a significant effect during the 3rd rocker for overground, which further increases for downslope gait.

5.5 Conclusion

Findings indicate that ankle bracing in restricted mode has lessened ankle involvement throughout the stance phase rockers in both gait modes. The compensations to restricted ankle at the 1st rocker occur primarily in the involved knee joint as a result of weight acceptance by the increase of knee flexion loading response which was likely due to delay of attainment foot-flat on the ground. The knee joint compensated that having shown an increase of positive work in the 2nd rocker to return the limb to an optimal length to counter increased knee flexion loading response. For a period of the 3rd rocker phase, involved knee increases negative work during downslope compared overground gait but reduces with the restricted ankle. This suggests that to control body transition in both gait modes, participants with restricted ankle compensated primarily at the knee joint. As expected, the combined variable limb negative/positive rotational work reflects surface and ankle conditions. It can, therefore, be assumed that the individuals that employ ankle bracing and/or a unilateral trans-tibial amputation with rigid ‘ankle’ prosthetic device should be able to walk down slopes up to 5 degrees as competently and safely as when walking overground. Nevertheless, the increased knee involvement has to be taken into consideration. This highlights that the restricted ankle has a mostly similar effect in both gait modes.

CHAPTER SIX - BODY DYNAMICS: MICROPROCESSOR CONTROLLED HYDRAULICALLY DAMPED PROSTHETIC VERSUS CONVENTIONAL ANKLE DURING RAMP DESCENT IN UNILATERAL TRANS-TIBIAL AMPUTEES

6.1 Introduction

Lower-limb amputees have to remodel their gait pattern to correspond to their prosthesis functionality and environment. Basic prosthetic foot devices have a rigid/non-articulated ‘ankle’ that provides good stability during weight bearing. Rigid prosthetic foot devices do not provide articulation and function depends on the deformation and recoil properties of the heel and fore-foot keel. The gait cycle is divided into three sub-sequential rockers. To attain foot-flat with the dynamic response prosthetic foot, a pseudo ‘plantar-flexion’ in the 1st rocker is required and could be achieved only through deformation of the prosthetic heel which leads to foot-flat on the ground and provides a stable position within weight transfer over the supporting limb. In the subsequent 2nd rocker, the support limb shank rotates and transfers body weight throughout the single-limb-support (SLS) which requires control of the body dynamics. The 2nd rocker was proposed by Cavagna et al as a pendulum model (Cavagna et al. 1963; Cavagna and Margaria 1966) to describe the high efficiency of the human bipedal locomotion. In the 3rd rocker ‘dorsi-flexion’ is attained due to the deformation of the fore-foot keel which is followed by its recoil (Perry et al. 1992). There the 1st and 3rd rockers in the pendulum model were described as absorption/redirection and propulsion accordingly between SLS phase.

Improved prosthetic devices have an articulated ‘ankle’ unit with dynamic-response heel and fore-foot keel components. The benefit of an articulated ankle-foot device is to prolong the stance time which improves the symmetry of the gait as TT’s have reduced stance time on the prosthetic side. Early researchers demonstrated that articulated ankle-feet devices would benefit TT gait compared to the rigid ankle in overground gait (Nolan et al. 2003; Zmitrewicz et al. 2007). Later studies have indicated the utilisation of the dynamic response feet with a hydraulically articulated ‘ankle’ (*Echelon*; Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) provides biomechanical benefits for overground gait in TTs (Portnoy et al. 2012; De Asha et al. 2013a; De Asha et al. 2014). However, the functionality of non-adaptable prosthetic ankle-foot devices would be dependent on the ‘rubber-snubber’ properties of elastic-AF

(*Epirus*; Chas. A. Blatchford and Sons Ltd., Basingstoke, UK)) or hydraulic flow nonMC-AF (*Echelon*; Chas. A. Blatchford and Sons Ltd., Basingstoke, UK)) as well as the stiffness of the e-Carbon spring control of plantar/dorsi-flexion. The downside of this design is that the articulated ankle-foot devices are set-up for overground gait at self-selected walking speed so do not adapt to patients walking velocity or the slope ambulation (Vrieling et al. 2008).

Daily activities commonly include slope ambulation and require adaptation of the locomotive pattern distinct to overground gait (Lay et al. 2006; McIntosh et al. 2006). The evaluation of ramp descent examines the higher risk of falling due to anterior-posterior instability in contrast to overground gait (McFadyen and Winter 1988; Fraser et al. 2007). Gait down the slope compared to overground requires an increase of control strategies and has to be a compromise between minimising energy consumption and maintaining stability (Hunter et al. 2010). Slope ambulation is known as a very challenging task for TT's (Macfarlane et al. 1991; Sin et al. 2001), predominantly descent requires increased control of body weight (Vickers et al. 2008). Therefore, during the stance phase, ankle articulation has a crucial role in maintaining dynamic stability (Vickers et al. 2008). TT's have to adapt their body dynamics according to the approaching terrain and the functionality of the ankle-foot device to minimise the risk of loss of balance which may result in a fall or slip.

Adaptive prosthetic ankle-foot devices were designed to function according to the approaching terrain. The role of the adaptive prosthesis in upslope gait is to minimise effort but to control body motion down the slope; it is critical due to increased potential gravitational energy. Ability to control body dynamics is crucial to attaining foot-flat sooner as it delivers a stable position within weight transfer on the supporting limb (Perry et al. 1992). The residual limb knee must flex to achieve foot-flat during slope descent with a non-articulating ankle-foot device, which creates an external knee flexion moment that accompanying torque at the residuum/socket interface (Vickers et al. 2008). Therefore to maintain the knee in a flexed position requires compensatory control from the thigh muscle. Otherwise, it would increase body motion, which would affect the amputee's stability during ramp descent. Adaptive, articulating ankle-foot devices are intended to overcome this, thereby improving dynamic stability and reducing the effort for users.

The microprocessor controlled quasi-passive hydraulic ankle-foot device *Élan* (MC-AF) (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) is a newly commercially available ankle-foot device that claims to improve amputees' gait pattern on inclined surfaces. However, a full insight of body dynamics is unknown. Therefore, it is required to certify that *Elan* provides optimal functionality throughout the stance phase while gaining deeper insights that will help amputees' rehabilitation. The *Elan* ankle-foot claims to be able to adapt accordingly to speed and terrain. During ramp descent, an adaptive mode of *Elan* has reduced 'plantar-flexion' resistance of hydraulic flow to achieve foot-flat quicker. This follows the body transfer throughout the SLS, where the *Elan* hydraulic flow increased resistance 'dorsi-flexion' to provide safe and controlled body transfer over a support foot. This change in functionality should improve dynamic stability during ramp descent for TT.

Interestingly, the research of Fradet et al. questioned the benefits of adaptive ankle-foot Proprio-Foot (Ossur hf, Iceland) on ramp descents (Fradet et al. 2010). However, the patients indicated reduced stress on the knee joint and feeling safer during ramp descent (Fradet et al., 2010). It has also been noticed, that *Elan* from this study functions during the stance phase, but the Proprio-Foot adapts only in the swing phase. During stance phase, it acts as a conventional dynamic response foot with rigid 'ankle' (Au et al. 2007a; Versluys et al. 2008; Eilenberg et al. 2010). Therefore, investigations of an adaptive ankle-foot prosthetic device that can adapt to different terrains during stance phase are required.

This study investigated the effect of different types of prosthetic ankle-foot articulations on body dynamics during ramp descent with two different walking speeds in TTs. Currently; there were no known studies that include the effect of the ankle-foot prosthetic devices with different types of articulation on body dynamics during down the ramp gait in TT patients. A comparison was performed between articulating ankle-foot devices: *Epirus* (elastic-AF) and *Elan* (in active mode (MC-AF) and non-active mode (nonMC-AF)). To assess the dynamic stability of TT during SLS, a variable Virtual Limb of the support limb and CoM. The study's investigation of VL behaviour will provide deeper insight into how ankle-foot devices affect body dynamics in down the ramp gait. This would display changes of body weight motion relative to the supporting foot, according to the ankle-foot device functionality. Therefore, the VL variable displays behaviour of CoM motion relative to the support foot during a stance phase, which can be contrasted to the Centre-of-Pressure (CoP) that shows an effect of the body progression (line of ground reaction forces (GRF) action) on the ground that also depends on ankle motion/function. CoP displacement occurred mainly underneath the prosthetic ankle-foot. The study would link both parameters to provide a clearer picture of the ankle-foot device's functionality. The assessment of CoM velocity, VL angular velocity and CoP velocity for overground and ramp descent gait could have a different statistical outcome between prosthetic ankle-foot devices. The evaluation of those variables would clarify the effect of the prosthetic ankle-foot device on the pendulum model as the pendulum behaviour would depend on gait mode and ankle-foot articulation.

The purpose of the study is to investigate body dynamic alterations between adaptive (MC-AF), non-adaptive hydraulic (nonMC-AF) and elastic (elastic-AF) prosthetic 'ankle' articulations during ramp descent in TT. The main hypothesis of the study, the VL angular velocity during ramp descent would be increased during SLS for the non-adaptive (elastic-AF or nonMC-AF) compared to adaptive (MC-AF) device due to increase of gravitational potential energy (Chapman 2008) as a result of uniform articulation so the inability to attain foot-flat sooner and rotate over the support foot slower. Alternatively, ramp descent of TTs with non-adaptive (elastic-AF or nonMC-AF) ankle-foot prosthetic devices would have increased knee flexion and increase the support shank angular velocity compared to adaptive (MC-AF). In another hand, CoP velocity during SLS would be increased for non-adaptive (elastic-AF or nonMC-AF) as a result of the inability of ankle-foot to change articulation resistance according to increased gravitational energy so confirm the improvement of dynamic instability. Another question of this research was if the use of MC-AF ankle-foot articulation would improve symmetry of ramp descent in TTs. This would be as result of an increase in control of prosthetic ankle-foot articulation so the controlled transition of TTs would be performed more symmetrical between the intact ankle and prosthetic ankle-foot.

6.2 Methods

6.2.1 Participants and Ethics

Nine physically active, males with a unilateral trans-tibial amputation (mean (SD) age 41.2 (12.9) years, mass 74.14 (15.7) kg, height 1.76 (0.06) m), participated in this study. All participants' amputees were classed as at least K3 on the Medicare scale. Full details are presented in chapter 3.4.2. Ethical approval for this study was conducted in accordance with the tenets of the Declaration of Helsinki and granted from the University of Bradford's Committee for Ethics in Research (ref. number E.119).

6.2.2 Specific equipment, procedure, data acquisition and processing

All amputee participants were familiarised with each prosthetic device (*Epirus, Elan*) described in chapter 3.12. Details of each prosthetic device (*Epirus, Elan*) with full walking protocol are presented in chapter 3.12.

Six successful trials were completed (landing precisely on of the force platform without gait alterations) for prosthetic and intact limb in each prosthetic condition in counterbalance order among the participants. Gait down the ramp was assessed in two blocks in counterbalanced order across participants. In one block participants used an *Epirus* and in the other they used an *Elan*. Participants were instructed to walk the first set at self-selected walking speed as they would normally walk down the ramp and the second set at comfortable slow speed. The trials of the block with the *Elan* device were undertaken in random order, where the microprocessor being manipulated in active (MC-AF) or inactive (nonMC-AF) mode. The device in non-active (nonMC-AF) mode behaves just like the *Echelon* (Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) ankle-foot device. The order the between active (MC-AF) and inactive (nonMC-AF)) modes of *Elan* device was manipulated ‘blindly’ for participants. The manipulation was performed remotely at the start of a trial via Bluetooth connection with the foot’s microprocessor.

To examine down the ramp gait, a custom made modular ramp 2.8 metres long with an inclination of 5 degrees and 1.0-metre long level ground landing was used. Full details are provided in chapter 3.7.6. Full details of all the laboratory equipment used, how kinetic and kinematic data were recorded and the full protocol procedure is presented in chapter three. Passive retro-reflective markers identified participants according to six degrees of freedom (6 DoF) model also described in chapter three.

The detailed description of data recording, processing and filtering provided in chapter three.

The stance phase was defined from initial contact (IC) till toe-off (TO) and verified from vertical components (Z) of ground reaction forces with a threshold of 20 N. The single-limb-support (SLS) was defined by kinematic data with the stance phase from TO till the IC of the contralateral foot. TO was created according to Zeni gait event detection (Zeni Jr et al. 2008). IC was determined at the threshold of the heel virtual marker's vertical velocity reduced below 0.15 m/s.

6.2.3 Data analysis

The variables were determined within Microsoft Excel (Microsoft, New York, NY, USA) during stance phase on the prosthetic side for each trial and then averaged across trials to provide the main parameter for each condition per participant. The following variables were assessed and listed below. Prosthetic side: knee loading response flexion was defined as maximum knee flexion during early stance (deg.); knee single-limb-support minimum flexion was defined as minimum knee flexion during single-limb-support phase (deg.); shank/pylon mean angular velocity during single-limb-support was defined as a shank/pylon rate of rotation forward in the sagittal plane within the global coordinate system (˚sec-1); the timing of attaining foot-flat (s.) was defined from IC to prosthetic foot virtual toe marker velocity drop below 0.1 ms^{-1}; limb stance phase forward CoM mean (A-P) velocity; VL mean angular velocity during SLS (˚sec^{-1}); VL and shank angle at foot-flat (˚); step length was defined from the virtual heel marker of the lead foot till toe virtual marker of trail foot at IC; stance time. Anterior-posterior CoP velocity at foot-flat, mean from IC to foot-flat, mean during SLS. The first 5 ms of CoP data were eliminated to avoid any scuffs on the surface in CoP data. The angular velocity of the prosthetic side VL and/or shank during SLS and was defined as the rate of rotation of the segment in the sagittal plane within the global coordinate system (˚sec^{-1}). Intact side: step length, stance time, knee flexion loading response; CoM forward velocity throughout the stance phase.

The index of symmetry (IOS) was calculated for step length, stance time, and limb stance phase forward CoM mean (A-P) velocity. The calculation of SI index was firstly presented by Robinson et al. (1987) and adapted for unilateral amputees by Nolan et al. (2003) (Robinson et al. 1987; Nolan et al. 2003) as presented below.

$$SI = ((X_{intact} - X_{prost}) / 0.5(X_{intact} + X_{prost})) \times 100\%,$$

where X_{intact} is a variable from the intact side and X_{prost} is the corresponding variable from the prosthetic side. There, a positive value indicates greater magnitude on the intact side and a negative value indicates greater magnitude on the prosthetic side. Parameters were calculated for each individual trial, then averaged across trials to give a mean value for each prosthetic and walking condition per participant.

6.2.4 Statistics

To determine differences between ankle-foot articulations (MC-AF, nonMC-AF and elastic-AF) and gait modes (comfortable slow and self-selected) repeated measures of ANOVA were used. Effect size differences (low $d < 0.3$, moderate $0.3 < d < 0.5$ and high $d < 0.5$) were calculated as Cohen's (Cohen 1988). Statistical analyses were performed in Statistica (v6, StatSoft, Inc., Tulsa, OK, USA). To identify any significance between conditions a post hoc comparison with Turkey HSD tests was used. The level of significance was set $p < 0.05$.

6.3 Results

6.3.1 Residual limb

TTs' residual side VL angular displacements during ramp descent (SSWS and comfortable slow) with elastic-AF, nonMC-AF and MC-AF articulations and able bodied individuals' involved limb during overground and ramp descent gait with restrictricted and non-restricted ankle condition are illustrated in Figure 35. Figure 33 illustrates the example of the same participant CoP forward velocity for self-selected and slow walking speed. The mean (±SD) and statistical significance of CoM velocity, VL and shank angular velocity during the stance on residual limb with elastic-AF, nonMC-AF and MC-AF articulations ramp descent for self-selected and slow walking speed is presented in Table 8. Table 10 illustrates the parameters: step length, stance time. Mean loading response peak knee flexion is illustrated in Figure 34. There were no significant interactions between the speed of ambulation and prosthetic ‘ankle’ articulation, so this will not be presented further in the results section unless stated otherwise. Effect size (d) between MC-AF and nonMC-AF or elastic-AF articulations are presented (*in italics*) in all Tables below (8-12).

There were main effects of walking speed ($F_{(1,8)}$=5.32, $p = 0.034$) and ankle-foot device type ($F_{(2,16)}$=7.67, $p < 0.001$) on the VL angle at the foot-flat. The VL angle at the foot-flat was acutest at the slow walking speed, and irrespective of speed was acuter for MC-AF and nonMC-AF compared to the elastic-AF ($p > 0.039$) between each ankle-foot articulation type (Table 8). There was a main effect of walking speed ($F_{(1,8)}$=55.30, $p < 0.001$) but not ankle-foot articulation type ($F_{(2,16)}$=0.98, $p = 0.40$) on the VL mean angular velocity during SLS (Table 8). The VL mean angular velocity during SLS was higher for self-selected walking speed (Figure 35). There were main effects of speed ($F_{(1,8)}$=28.63, $p < 0.001$) ankle-foot type ($F_{(2,16)}$=7.23, $p = 0.006$) on shank angular velocity during SLS. Shank angular velocity during SLS was slower at the slow walking speed, and irrespective of speed was significantly slower for the MC-AF than either the nonMC-AF or elastic-AF, with no significant differences between the nonMC-AF and elastic-AF (Table 8).

Table 8. Residual side group mean (±SD) of prosthetic limb VL angles at foot-flat; VL angular velocity for the period of the single-limb-support (SLS) with elastic-AF, nonMC-AF and MC-AF prosthetic devices in ramp descent gait with self-selected and slow walking speed. Where differences between MC-AF and nonMC-AF or elastic-AF articulations are effect sizes Cohen's (d) presented (in *italics*).

	Ramp Descent Slow			***Ramp Descent SSWS***			p *value* (*F value*)
	MC-AF	*nonMC-AF*	*elastic-AF*	*MC-AF*	*nonMC-AF*	*elastic-AF*	
VL angle at foot-flat (˚)	4.1 (1.6)	3.1 (1.5) *0.7*	4.9 (1.5) *0.5*	4.1 (1.8)	3.4 (1.4) *0.5*	5.8 (1.5) *0.9*	**Speed 0.034 (5.32)** **Foot 0.001 (7.67)** Int. 0.12 (2.21)
Mean VL angular velocity SLS ($˚sec^{-1}$)	56.7 (5.7)	58.4 (6.7) *0.3*	56.6 (3.7) *<0.1*	77.6 (8.3)	78.7 (7.8) *0.1*	77.2 (9.7) *<0.1*	**Speed<0.001 (55.30)** Foot 0.40 (0.98) Int. 0.92 (0.06)
Shank angle at foot-flat (˚)	3.0 (3.8)	5.5 (3.6) *0.7*	1.6 (4.4) *0.3*	4.9 (3.8)	6.8 (3.7) *0.5*	2.0 (5.2) *0.6*	Speed 0.44 (0.65) **Foot 0.023 (4.80)** Int. 0.23 (1.59)
Mean Shank angular velocity SLS ($˚sec^{-1}$)	63.8 (12.5)	70.4 (14.5) *0.5*	70.8 (11.4) *0.6*	85.1 (15.6)	91.2 (14.6) *0.4*	92.0 (19.9) *0.4*	**Speed<0.001 (28.63)** **Foot 0.006 (7.23)** Int. 0.96 (0.05)

Time to foot-flat was significantly longer at the slow walking speed ($F_{(1,8)}$=34.93, p < 0.001). Time to foot-flat was significant between ankle-foot articulation types ($F_{(2,16)}$=37.82, p < 0.001), irrespective of speed, time to foot-flat was shortest for the elastic-AF, than MC-AF and longest for the nonMC-AF articulation with the differences between each ankle articulation type being significant (Table 10). There were main effects of walking speed ($F_{(1,8)}$=15.73, p=0.006) but unaffected by ankle-foot device type ($F_{(2,16)}$=1.26, p = 0.31) on the timing of foot-flat kept on the ground (% stance). Timing of foot-flat on the ground (% stance) was lengthier at the slow walking speed. There was a significant speed by ankle-foot type interaction ($F_{(2,16)}$=5.68, p = 0.012). The interaction indicated that timing of foot-flat on the ground (% stance) was significantly lengthier for the MC-AF compared to nonMC-AF and elastic-AF devices at the slow walking speed, while at the self-selected walking speed, there was no effect between the ankle-foot articulation types.

Table 9. Residual side group mean (±SD) timing to Foot-flat; the timing of heel off relative to the stance (%) between elastic-AF, nonMC-AF and nonMC-AF prosthetic devices in ramp descent gait with self-selected and slow walking speed. Where differences between MC-AF and nonMC-AF or elastic-AF articulations are effect sizes Cohen's (d) presented (in *italics*).

	Ramp Descent Slow			***Ramp Descent SSWS***			p *value* *(F value)*
	MC-AF	*nonMC-AF*	*elastic-AF*	*MC-AF*	*nonMC-AF*	*elastic-AF*	
Time to foot-flat (s)	0.195 (0.016)	0.212 (0.026) *0.7*	0.170 (0.019) *1*	1.172 (0.016)	0.180 (0.020) *0.4*	0.150 (0.021) *1*	**Speed<0.001 (34.93)** **Pros.< 0.001 (37.82)** Int. 0.058 (3.42)
Timing of Foot-flat on the ground (% stance)	39.0 (5.5)	34.8 (8.2) *0.5*	33.8 (7.8) *0.7*	31.4 (8.1)	31.2 (9.2) *<0.1*	30.3 (9.8) *0.1*	**Speed 0.006 (15.73)** Foot. 0.31 (1.26) **Int. 0.012 (5.68)**

There were main effects of speed ($F_{(1,8)}$=10.12, p =0.011) and ankle-foot articulation type ($F_{(2,16)}$=4.02, p =0.036) on residual-knee peak loading response flexion (% stance) (Table 9). The knee peak occurred sooner at self-selected walking speed, and irrespective of speed was significantly sooner when using the MC-AF than elastic-AF (p = 0.030). However, there was a significant speed by ankle-foot type interaction ($F_{(2,16)}$=5.21, p = 0.020). This interaction indicated that knee peak loading response flexion (% stance) was occurring sooner for the MC-AF compared to nonMC-AF and elastic-AF devices at the slow walking speed, while at the self-selected walking speed, knee peak flexion (% stance) was sooner for the MC-AF but only compared to the elastic-AF device. There were main effects of speed ($F_{(1,8)}$=36.45, $p < 0.001$) and ankle-foot type ($F_{(2,16)}$=3.98, $p = 0.039$) on knee peak loading response flexion (Table 9). The knee peak loading response flexion was increased at self-selected walking speed, and irrespective of walking speed was significantly reduced when using the MC-AF than nonMC-AF or elastic-AF devices ($p = 0.039$). There was the main effect ankle-foot type ($F_{(2,16)}$=8.89, $p = 0.003$) but not on walking speed ($F_{(2,16)}$=0.53, $p = 0.49$) on single-limb-support mean residual-knee flexion (Table 10). The mean knee flexion was less flexed irrespective of walking speed; it was significantly less flexed when using the MC-AF than nonMC-AF or elastic-AF devices. However, there was a significant speed by ankle-foot type interaction ($F_{(2,16)}$=4.48, $p = 0.029$). This interaction indicated that mean residual-knee flexion was significantly less with the MC-AF compared to nonMC-AF and even less compared to elastic-AF ankle-foot devices at the slow walking speed, while at the self-selected walking speed, was significantly less with the MC-AF compared to nonMC-AF or elastic-AF the ankle-foot devices. Centre-of-Mass mean velocity throughout the stance has the main effect of walking speed ($F_{(1,8)}$=40.01, $p < 0.001$) which was slower at the slow walking speed without the effect of ankle-foot device type ($F_{(2,16)}$=1.16, $p = 0.28$) (Table 9). Shank angular velocity (Figure 34) during single-limb-support (Table 12) was significantly lower at the slow walking speed ($F_{(1,8)}$=28.63, $p < 0.001$). Shank, angular velocity was significant between ankle types ($F_{(2,16)}$=7.23, $p = 0.006$), irrespective of speed, was significantly lower for the MC-AF articulation than either the nonMC-AF or elastic-AF articulation, with no significant differences between the nonMC-AF and elastic-AF articulations. Loading response flexion was significantly reduced at the slow walking speed ($F_{(1,8)}$=36.45, $p < 0.001$). Loading response flexion was significant between ankle articulation types ($F_{(2,16)}$=3.98, p = 0.039) (Table 12), and irrespective of speed, was significantly reduced when using the MC-AF foot than either the nonMC-AF or elastic-AF articulation (Figure 34). Single-limb-support minimum flexion was significantly increased at the slow walking speed ($F_{(1,8)}$=3.53, p = 0.024). Single-limb-support minimum flexion was significant between ankle articulation types ($F_{(2,16)}$=12.89, p < 0.001) (Table 12), and irrespective of speed was significantly reduced when using the MC-AF than either the nonMC-AF or elastic-AF articulation (Figure 34). Knee loading response flexion or single-limb-support minimum flexion indicated no significant difference between the nonMC-AF and elastic-AF articulations (p = 0.77).

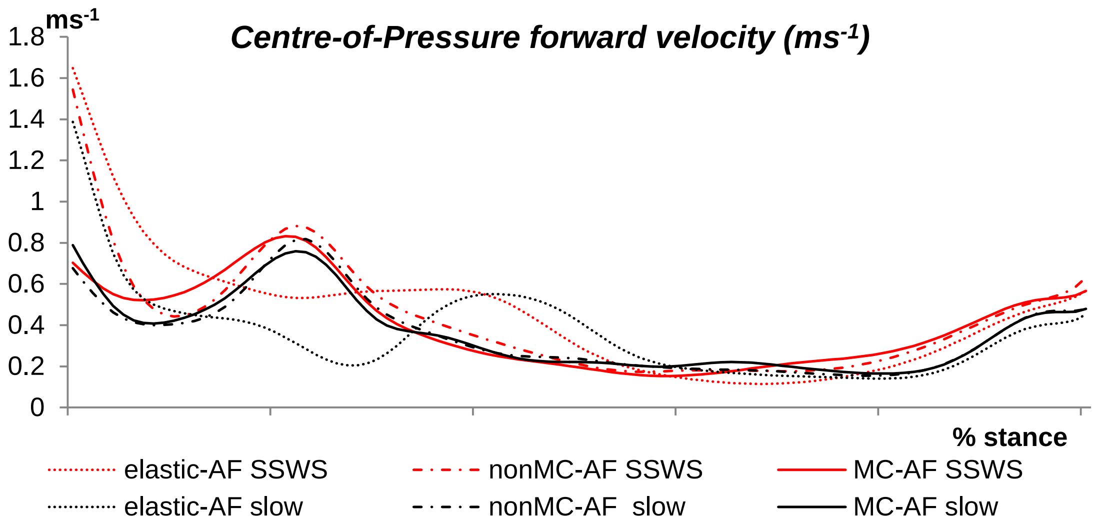

Figure 33. Exemplar of the same participant Centre-of-Pressure (CoP) forward velocity for self-selected walking speed (SSWS) (red) and slow (black) when using an MC-AF, nonMC-AF and elastic-AF. CoP forward velocity of the residual side is drawn to scale (MC-AF slow –ramp descent slow MC-AF; nonMC-AF slow – ramp descent slow nonMC-AF; elastic-AF slow – downslope slow elastic-AF; MC-AF SSWS – ramp descent, self-selected walking speed MC-AF; nonMC-AF SSWS– ramp descent self-selected walking speed nonMC-AF; elastic-AF SSWS– ramp descent self-selected walking speed elastic-AF).

Table 10. Residual side group mean (±SD) step length (m), stance time (m), knee loading response (deg.), knee peak loading response relative to the percentage of stance, Knee flexion mean during SLS (deg.), CoM velocity throughout the stance and for the period of the single-limb-support (SLS) (ms^{-1}) with MC-AF, nonMC-AF and elastic-AF prosthetic devices in ramp descent gait with self-selected and slow walking speed. Where differences between MC-AF and nonMC-AF or elastic-AF articulations are effect sizes Cohen's (d) presented (in *italics*).

	Ramp Descent Slow			***Ramp Descent SSWS***			p *value*
	MC-AF	*nonMC-AF*	*elastic-AF*	*MC-AF*	*nonMC-AF*	*elastic-AF*	*(F value)*
Step length (m)	0.57 (0.05)	0.58 (0.05) *<0.1*	0.57 (0.03) *0.1*	0.68 (0.05)	0.68 (0.04) *0.1*	0.67 (0.06) *0.1*	**Speed<0.001 (104.46)** Foot 0.22 (1.241) Int. 0.70 (0.36)
Stance time (s)	0.791 (0.076)	0.777 (0.081) *0.2*	0.782 (0.06) *0.1*	0.659 (0.063)	0.653 (0.060) *0.1*	0.658 (0.062) *<0.1*	**Speed<0.001 (26.61)** Foot 0.42 (0.59) Int. 0.67 (0.41)
Knee loading response (°)	18.0 (5.4)	21.3 (6.1) *0.6*	21.7 (5.1) *0.7*	21.8 (4.7)	24.7 (5.3) *0.6*	24.4 (5.3) *0.5*	**Speed <0.001 (36.45)** **Foot 0.039 (3.98)** Int. 0.57 (0.57)
Knee peak loading response (% of stance)	29.9 (3.6)	34.6 (3.4) *1*	38.0 (9.4) *1*	28.5 (2.7)	31.1 (2.6) *0.9*	32.0 (6.5) *0.7*	**Speed 0.011 (10.12)** **Foot 0.036 (4.02)** **Int. 0.020 (5.21)**
SLS minimum knee flexion (°)	10.8 (7.7)	16.7 (7.6) *0.6*	17.4 (6.7) *0.9*	7.1 (7.4)	12.1 (9.3) *0.7*	13.4 (6.2) *0.7*	**Speed 0.024 (3.53)** **Pros.< 0.001 (12.89)** Int. 0.94 (0.48)
CoM mean velocity throughout the stance (ms^{-1})	1.03 (0.12)	1.06 (0.15) *0.2*	1.03 (0.08) *<0.1*	1.39 (0.16)	1.41 (0.15) *0.1*	1.39 (0.15) *<0.1*	**Speed <0.001 (40.01)** Foot 0.28 (1.16) Int. 0.94 (0.06)
CoM mean velocity throughout the SLS (ms^{-1})	0.97 (0.12)	1.01 (0.14) *0.3*	0.98 (0.08) *0.1*	1.34 (0.16)	1.36 (0.15) *0.1*	1.33 (0.15) *<0.1*	**Speed <0.001 (59.72)** Foot 0.26 (1.48) Int. 0.83 (0.16)

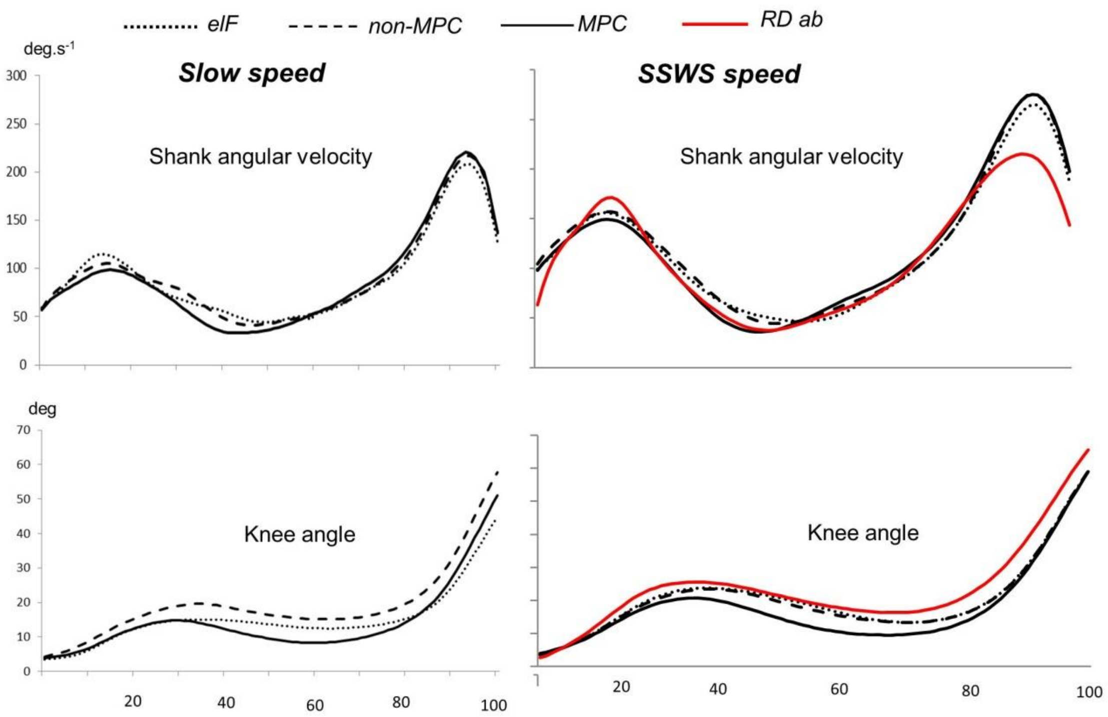

Figure 34 Ensemble group mean stance phase shank angular velocity and knee angular displacement when using the elastic-AF (black dotted line), nonMC-AF (black dashed line), MC-AF (black solid line) ankle-foot and RD ab -ramp descent able-bodied individuals (red solid line). Able-bodied data were obtained from chapter five.

There was a main effect of walking speed ($p < 0.001$) but not ankle-foot type ($p = 0.078$) on mean anterior-posterior CoP forward velocity during single-limb-support. However, there was a significant walking speed by ankle-foot type interaction ($p = 0.006$). The interaction has indicated, that mean CoP velocity was lower for the MC-AF compared to nonMC-AF and elastic-AF at the slow walking speed, while at the self-selected walking speed level, CoP velocity was lower for the MC-AF but only compared to the nonMC-AF ankle-foot device. There was a main effect of walking speed ($p = 0.038$) and ankle-foot device type ($p = 0.026$) on mean anterior-posterior CoP forward velocity from IC to foot-flat. Mean CoP velocity from IC to foot-flat has main effects of speed ($p < 0.001$) which was slower at the slow walking speed, there was irrespective of speed was significantly slower for the elastic-AF ankle-foot than either the nonMC-AF or MC-AF devices.

Table 11. Residual side group mean (±SD) Centre-of-Pressure (CoP) velocity at Foot-flat relative to the stance (%), mean from IC to foot-flat (ms^{-1}), mean during Single-limb-support (ms^{-1}) between elastic-AF, nonMC-AF and nonMC-AF prosthetic devices in ramp descent gait with self-selected and slow walking speed. Where differences between MC-AF and nonMC-AF or elastic-AF articulations are effect sizes Cohen's (d) presented (in *italics*).

	Ramp Descent Slow			***Ramp Descent SSWS***			p *value* *(F value)*
	MC-AF	*nonMC-AF*	*elastic-AF*	*MC-AF*	*nonMC-AF*	*elastic-AF*	
CoP velocity at foot-flat (ms^{-1})	0.869 (0.198)	0.982 (0.147) *0.7*	0.397 (0.065) *1.7*	1.037 (0.121)	1.096 (0.157) *0.6*	0.518 (0.052) *1.8*	**Speed<0.001 (27.88)** **Foot <0.001 (105.82)** Int. 0.43 (0.87)
CoP velocity mean from IC to foot-flat (ms^{-1})	0.442 (0.268)	0.495 (0.196) *0.1*	0.253 (0.182) *1*	0.597 (0.353)	0.589 (0.392) *0.2*	0.461 (0.352) *1*	**Speed 0.038 (6.19)** **Foot. 0.026 (4.65)** Int. 0.76 (0.28)
CoP velocity mean during SLS (ms^{-1})	0.282 (0.047)	0.307 (0.041) *0.6*	0.306 (0.033) *0.6*	0.358 (0.060)	0.372 (0.037) *0.3*	0,360 (0.035) *<0.1*	**Speed<0.001 (30.83)** Foot. 0.078 (3.01) **Int. 0.006 (7.28)**

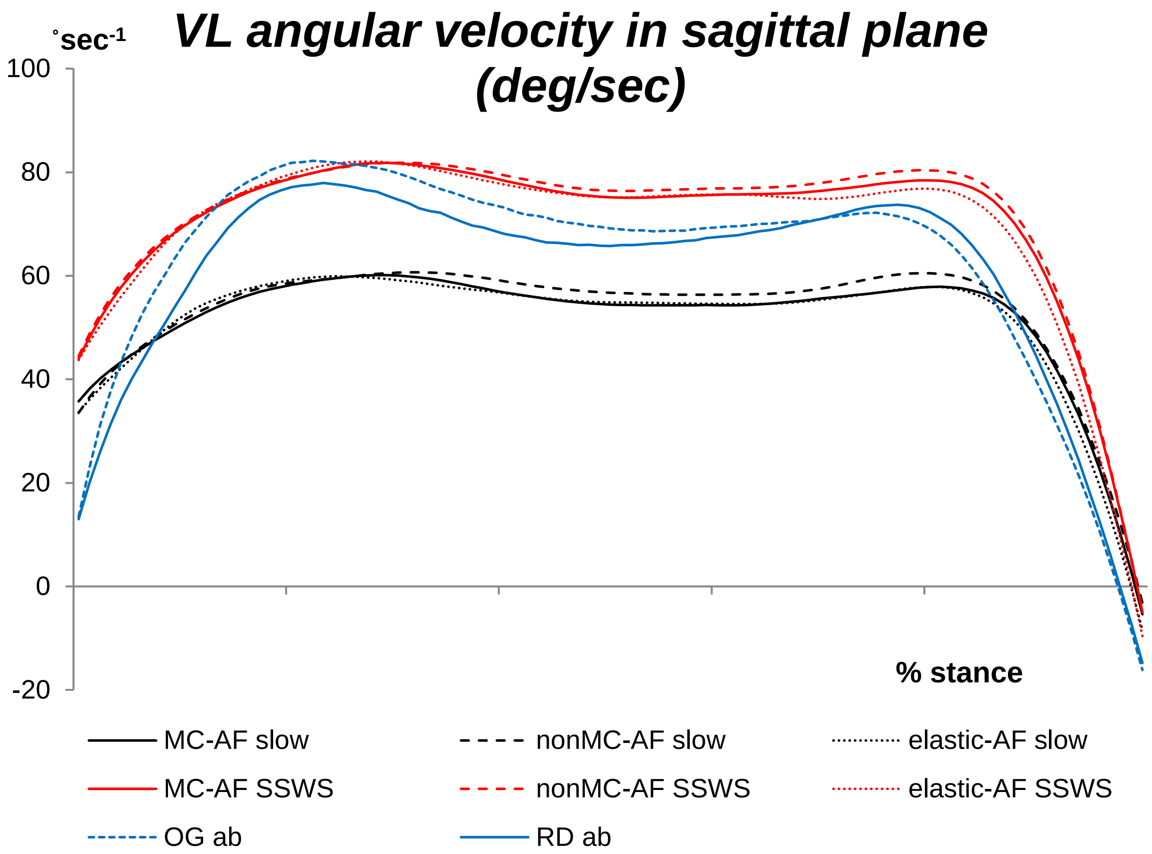

Figure 35. Residual side VL angular velocity (deg.s^{-1}) normalised to 100 points (stance phase), averaged across 9 TT participants. (MC-AF slow (black solid line); nonMC-AF slow (black dashed line); elastic-AF slow (black dotted line); MC-AF self-selected walking speed (red solid line); nonMC-AF self-selected walking speed (red dashed line); elastic-AF self-selected walking speed (red dotted line)). Blue lines (solid and dashed) are representing involved side VL angular velocity (deg.s^{-1}) normalised to 100 points (stance phase) and averaged across 20 able-bodied participants OG ab (blue dashed line) – overground able-bodied (ankle in non-restricted condition); RD ab (blue solid line) – ramp descent able-bodied (ankle in non-restricted condition) (Chapter four).

6.3.2 Intact side

The overall intact side results are summarised in table 12 below and figures plotted in Appendix 10. These following results are presented in the discussion below (Chapter 6.4).

Table 12 Intact side group mean (±SD) step length (m), stance time (m), knee loading response (deg.), CoM velocity throughout the stance (ms^{-1}) with MC-AF, nonMC-AF and elastic-AF prosthetic devices in ramp descent gait with self-selected and slow walking speed. Where differences between MC-AF and nonMC-AF or elastic-AF articulations are effect sizes Cohen's (d) presented (in *italics*).

	Ramp Descent Slow			***Ramp Descent SSWS***			p *value* (*F value*)
	MC-AF	*nonMC-AF*	*elastic-AF*	*MC-AF*	*nonMC-AF*	*elastic-AF*	
Step length (m)	0.57 (0.06)	0.58 (0.05) *0.3*	0.59 (0.05) *0.3*	0.68 (0.05)	0.68 (0.05) *0.1*	0.67 (0.05) *0.2*	**Speed<0.001 (32.67)** Foot 0.59 (0.55) Int. 0.31 (1.27)
Stance time (s)	0.804 (0.089)	0.797 (0.093) *0.1*	0.790 (0.085) *0.2*	0.674 (0.075)	0.676 (0.076) *0.1*	0.671 (0.061) *0.1*	**Speed<0.001 (27.18)** Foot 0.76 (0.28) Int. 0.79 (0.23)
Knee loading response (°)	22.5 (8.7)	23.5 (8.4) *0.1*	22.5 (6.3) *0.1*	28.6 (5.7)	29.0 (5.5) *0.1*	28.1 (4.8) *0.1*	**Speed <0.001 (27.42)** Foot 0.59 (0.54) Int. 0.85 (0.17)
CoM mean velocity throughout the stance (ms^{-1})	0.96 (0.15)	0.97 (0.14) *0.1*	0.97 (0.10) *0.1*	1.29 (0.17)	1.31 (0.16) *0.2*	1.25 (0.15) *0.2*	**Speed <0.001 (60.01)** Foot 0.12 (1.16) Int. 0.44 (0.46)

6.3.4 Symmetry

The SI examination of parameters did not inicate statistically significant differences between speeds and prosthetic ankle-foot articulations in step length (p = 0.37), stance time (p = 0.29) or CoM mean forward velocity during stance phase (p = 0.58).

6.4 Discussion

This study investigated what effects prosthetic ankle-foot articulations (adaptive (MC-AF), non-adaptive hydraulic (nonMC-AF) and elastic (elastic-AF)) would have during ramp descent with two different walking velocities on TT body dynamic. Returning to the hypothesis posed at the beginning of this section, it is now possible to state that during ramp descent foot-flat is lowered to the floor following initial contact quickest for elastic-AF ankle-foot then MC-AF ankle-foot articulation. The results of this investigation show that prosthetic ankle-foot articulation type did not have an effect on VL angular velocity during single-limb-support. However, mean forward anterior-posterior CoP velocity during single-limb-support indicated better dynamic stability with the MC-AF articulation. The absence of the effects of VL angular velocity during single-limb-support with different prosthetic articulations in down the ramp gait is likely to be the result of an increase in the mean knee flexion with an increased mean shank angular velocity during single-limb-support. This supports the study hypothesis and indicates that participants with MC-AF have increased dynamic stability during ramp descent. Also, results indicate that foot staying flat on the ground (relative to the percentage of stance) was maintained longer when participants used the MC-AF compared to when participants used the elastic-AF or nonMC-AF devices at slow speed. The discussion section of this chapter covers the study findings and gait data that was reported in the literature. Therefore, the synthesis of the results relating to body dynamics was discussed below.

Surprisingly, there were no known studies that have examined gait of TTs in a comfortable slow manner. However, TTs have reported that slow ramp descent requires more effort. The study results presented, ramp descent with slow compared to self-selected walking speed has the delay to establish a foot-flat ($p<0.001$). There was a reduced walking speed from self-selected to comfortable slow so as a result, the heel deformation within ‘ankle’ articulation was inadequate to attain foot-flat quicker. Slower time to attain foot-flat during ramp descent was lessens timing of body motion control which leads to instability (Perry et al. 1997; Vickers et al. 2008; Fradet et al. 2010). The attainment of foot-flat quicker establishes a stable base of support with anterior-posterior dynamic stability. There's no increase of load on the residuum during single-limb-support, however, shank/pylon mean angular velocity was reduced during single-limb-support and body weight transfer was slower. Furthermore, the stance time increased for comfortable slow speed, so participants were likely to sense a gradual increase of knee instability within the increased load on the residuum. Previously, researchers showed that TT had reduced stance time on the prosthetic side compared to intact (Engsberg et al. 1993; Nolan et al. 2003). Therefore, gait down the ramp has increased gravitational potential energy and delays to attain foot-flat, it will also delay when individuals could control this kinetic energy growth (Chapman 2008). So increased knee peak flexion loading response at slow ramp descent compared to self-selected walking speed ($p < 0.001$) is a result of increase of the kinetic energy growth control down the ramp which is supported by other researchers (Lay et al. 2006; McIntosh et al. 2006; Vickers et al. 2008; Vrieling et al. 2008). Another research indicated, in TT, the residual knee was being 'thrown/pushed' forwards in order to achieve foot-flat quicker (Vickers et al. 2008) which would also lead to knee peak flexion loading response increase. The increase of residual knee flexion loading response would likely lead to an increase of eccentric work which is known as partly amputates muscles and weakened (Winter and Sienko 1988). Despite this, no research has been found that examined slow ramp descent. Nevertheless, increased stance time within knee instability is likely to explain why ramp descent with slow speed is considered demanding for TTs. Hence, ramp descent with slow compared to self-selected walking speed could be presented as more demanding for TTs.

Gait down the ramp in TTs with a rigid 'ankle' prosthetic device delivers the sensation of 'pulling' the residual knee forward. The knee 'pulls' forward as compensation to attain foot-flat quicker (Vickers et al. 2008; Vrieling et al. 2008). Certainly, amputees with an articulated 'ankle' would attain foot-flat quicker, which would be a result not only of the heel deformity but also articulation at the 'ankle'. Earlier research presented that TT's have reported it was 'easier' to approach down the ramp with an articulated 'ankle' attachment compared to non-articulated (Su et al. 2010) and 'safer' with a prosthetic foot that can 'plantar-flex' during a swing phase (Fradet et al. 2010). Attainment of foot-flat is crucial during down the ramp gait for TT's; the study findings show that time to attain foot-flat was fastest when using the elastic-AF articulation and slowest when using the nonMC-AF articulation. Although, the foot-flat was attained faster when using the MC-AF compared to the nonMC-AF articulation ($p < 0.001$ with effect size moderate to high $d \geq 0.4$) and seemingly reduces knee peak loading response ($p = 0.039$ with effect size high $d=0.6$). A number of studies have already demonstrated the benefits of hydraulic 'ankle' in overground gait (De Asha et al. 2013a; De Asha et al. 2013b; Johnson et al. 2014). The finding of the present study suggests that the MC-AF ankle-foot articulation (ramp descent adaptive mode) has reduced hydraulic damping in ramp descent compared to conventional hydraulic nonMC-AF 'ankle'. Hence, the MC-AF (adaptive mode) ankle-foot articulation can 'plantar-flex' the prosthetic foot quicker than a conventional hydraulic 'ankle' which was set for overground gait. In the MC-AF 'ankle' with active ramp descent mode after the attainment of foot-flat followed the increase of 'dorsi-flexion' hydraulic dampening that control's shank forward rotation within body weight transfer over the support prosthetic foot. With the elastic-AF articulation attainment of foot-flat is performed through a combined mechanism, deformation of heel spring and 'rubber-snubber' ball hinge to 'plantar-flex' and attain foot-flat. However, after attainment of foot-flat, the 'rubber-snubber' ball joint would 'pull' the shank/pylon forward to the neutral position as a result of the elastic recoil properties of the 'rubber-snubber' hinge material. The neutral position is relative to the foot and pre-set by the prosthetist. The 'rubber-snubber' hinge recoils the shank/pylon motion towards 'dorsi-flexion'; this is restricted by 'hard stop' so 'dorsi-flexion' in an elastic-AF prosthetic device would be due to deformation of the fore-foot keel.

To deliver a stable base of support, the foot-flat has to stay flat on the ground longer. The foot stays flat on the ground (relative to the percentage of stance) longer with the MC-AF compared to the elastic-AF or nonMC-AF devices at slow speed (effect size moderate to high $d \geq 0.5$) for two different causes. If nonMC-AF ankle-foot attain foot-flat slower, but the elastic-AF device as a result of heel-off (heel rise) the ground due to the inability of ankle-foot mechanism to ‘dorsi-flex’ (‘dorsi-flexion’ is restricted by a ‘hard stop’) (Chapter 3.12). Prolonged uphold of foot-flat on the ground contributes to dynamic stability which importance increases on inclined surfaces (Vickers et al. 2008).

Surprisingly, no effects of articulation type were found on step length in ramp descent gait. On the other hand, the increase of the VL angle at foot-flat for elastic-AF or nonMC-AF compared to MC-AF articulation types was likely a result of the delay in knee peak flexion loading response (relative to % of stance) ($p = 0.036$ with effect size high $d \geq 0.7$) so participants would pull CoM further over the support limb (reduce angle). This has indicated that delay at foot-flat pulls CoM forward, which could have an effect on dynamic stability during ramp descent. This finding supports previous research of McIntosh et al. (McIntosh et al. 2006). The use of the MC-AF compared to other articulation types would enhance body motion control over the support limb during down the ramp gait which partially supports the second hypothesis of this chapter.

Another important finding was that the use of MC-AF ankle-foot device did not have an effect on mean VL angular velocity during single-limb-support ($p = 0.40$). The VL motion did not change between articulated ankle-foot devices and was likely a result of the increased mean knee flexion with non-adaptive throughout the single-limb-support. This also collaborates with a first experimental chapter (Chapter four) where restriction of the ankle in ankle-foot orthosis did not affect VL angular velocity during single-limb support within compensation at the knee joint. The figure 35 presents that amputee participants have higher self-selected walking speed and the VL angular velocity during single-limb-support than able-bodied individuals (Chapter four). These results are consistent with those of other studies and suggest that ramp descent is a demanding task for TT’s due to an increased requirement of body motion control (Sin et al. 2001; Vickers et al. 2008). Increased body transition in

TT has likely indicated that TT has a problem to control the transition of the body within increased potential gravitational energy. In contrast, TT have slower walking speed in overground gait (Nolan et al. 2003). Taken together, it could be suggested that TT has reduced capability to control body transition during ramp descent. The study presented, that occurrence of knee peak loading response relative to the percentage of the stance of stance was delayed for nonMC-AF and elastic-AF compared to MC-AF this was likely caused by the increase of mean knee flexion during the single-limb-support. It seems that ankle-foot articulation and the knee joint flexion are contributing to the control of pendulum transition over support foot when gravitational energy increases. To control increased gravitational potential energy growth, the ankle and knee joints absorb and redirect the body over support limb (Lay et al. 2007). However, elimination of the appropriate functionality of the ankle to approaching terrain would lead to compensation from the remaining lower-limb joints (Winter et al. 1990). The gravity-assisted body transfer over the support limb throughout single-limb-support phase, which leads to increased knee flexion compared to overground gait (Lay et al. 2006; McIntosh et al. 2006). There increase of knee flexion also contributed by the ineffective functionality of the non-adaptive ankle-foot devices. However, to enhance dynamic stability in ramp descent require an increase body transition control at the knee joint (Hunter et al. 2010). The adaptive function of the MC-AF ankle-foot device for down the ramp gait is to increase ‘dorsi-flexion’ resistance. The study results present that support knee flexion reduces with increased ‘dorsi-flexion’ resistance of MC-AF during single-limb-support. The increase of ‘dorsi-flexion’ resistance as a result of the increased demand of body motion control forward and downward during down the ramp gait. Subsequently, after the foot-flat phase, the function of the knee continues to be involved in controlled strategy for a downward and forward transition that aided by gravitational potential energy. The result is on the lines of earlier literature (Vickers et al. 2008) that found participants with a rigid ‘ankle’ have increased knee flexion compared to able-bodied individuals during mid-stance. Moreover, also corroborates with the findings of a great deal of the previous work in this field, there in able-bodied individuals knee flexion aided sagittal plane rotation of the tibia that ‘pulls’ body forward over the stance foot (Lay et al. 2006; McIntosh et al. 2006). This confirms the idea that gravitational energy assists fall from the contralateral limb where the function of the ankle and knee is to absorb and redirect this energy. It was also suggested that the prosthetic foot design should favour the control of the knee flexion within stability.

Results of mean CoP forward velocity during single-limb-support indicated that the MC-AF compared to nonMC-AF and elastic-AF ankle-foot articulations at the slow speed level was slower; whereas at the self-selected walking speed was slower only compared to the nonMC-AF articulation. This is most likely as a result of amputee participants have a problem controlling dynamic stability with non-adaptive prosthetic feet (nonMC-AF and elastic-AF) during slow speed which also supported by amputees report. The increased hydraulic flow resistance in MC-AF during single-limb-support increases dynamic control down the ramp gait. Although, the amputee participants at the self-selected walking speed have a slower CoP velocity with MC-AF device, but surprisingly also slower with elastic-AF compared to the nonMC-AF prosthetic device. This is likely as a result of amputees with elastic-AF ankle-foot to attain foot-flat quicker relative to the percentage of stance, so participants were able to control CoP progression. Interestingly, different foot prosthetic devices do not have an effect on mean CoP velocity during single-limb-support in overground gait (De Asha et al. 2013a) where there was, more importantly, smoothness of progression. This was not surprising as a function of the ankle-foot in overground is to act as a fulcrum in the pendulum model. Nevertheless, the function of the ankle-foot has changed for the down the ramp gait where control of the body transition was crucial for dynamic stability. On the other hand, CoP velocity progression reflects body CoM transferred through the prosthetic foot device (De Asha et al. 2013a).

Although the examination effects of ankle-foot articulation during the ramp descent on spatio-temporal parameter symmetry not presented in the current study, previous research presented effects of prosthetic ankle function on the residual side (Vickers et al. 2008). One possible explanation for the little differences between distinct ankle-foot articulations that experienced amputee participants may have tried to maintain limb symmetry. But further examination of the intact side would present additional insights of ankle-foot articulation effects, as ramp descent has greater demand on lower-limbs than overground gait (McIntosh et al. 2006; Lay et al. 2007; Franz et al. 2012). Hence, inapt prosthetic ankle-foot articulation in this challenging task could potentially lead to overload on the intact side within the increase of asymmetry in musculoskeletal function. Vickers et al. found greater support time on the intact compared to prosthetic side (Vickers et al. 2008), and Agrawal et al. suggested that enhanced functionality prosthetic feet improve symmetry between limbs during ramp descent (Agrawal et al. 2015). It can thus be suggested that improved prosthetic design could have positive effects on the intact limb during ramp descent.

The findings suggest MC-AF compared to other two ankle-foot articulations have reduced residual-knee loading response flexion which was likely to be driven by a combination of reduced ‘plantar-flexion’ resistance, and followed increased ‘dorsi-flexion’ resistance. Although, the use of different ankle-foot articulations did not affect VL angular velocity in single-limb-support. The most likely explanation of the result is that amputee participants with non-adaptive (elastic-AF or nonMC-AF) compared to adaptive (MC-AF) ankle-foot articulations have increased mean knee flexion with increased shank angular velocity. This supports the section hypothesis and indicates that use of MC-AF participants has enhanced dynamic stability during ramp descent. Also, the CoP forward velocity during single-limb-support would be reduced for an MC-AF prosthetic device so confirms the improvement of amputees’ dynamic stability. While the previous studies examined the effects of prosthetic devices on the CoP and/or the whole body CoM transition would be beneficial to employ the VL parameter within its joints to gain more profound insights into ankle-foot articulations. Future work should investigate further biomechanical differences between ankle-foot articulations during ramp descent.

6.5 Conclusion

The investigation has demonstrated, the VL angular velocity during single-limb-support in the sagittal plane was not affected by articulation type, so whole body pendulum motion down the ramp does not depend on the prosthetic attachment. However, the knee joint has reduced contribution for adaptive compared to non-adaptive ankle-foot devices into the control of pendulum transition down the ramp. Participants with the MC-AF ankle-foot articulation attained foot-flat second fastest (after elastic-AF), maintained foot-flat on the ground longer, and had a reduction of the knee flexion. These were considered an improvement in dynamic stability during ramp descent. Reduced CoP velocity during single-limb-support was also indicated an improvement of dynamic stability with adaptable (MC-AF) ankle-foot articulation compared to non-adaptable (nonMC-AF and elastic-AF). There, it might perhaps be significant to suggest that elastic ankle-foot articulation was more responsive on surfaces with up to 5 degrees of inclination to attain foot-flat quicker. However, the elastic articulation does not provide controlled body weight transfer over the support limb.

CHAPTER SEVEN - EFFECT OF MICROPROCESSOR CONTROLLED ANKLE-FOOT COMPARED TO CONVENTIONAL ANKLE-FOOT ARTICULATION MECHANISMS DURING RAMP DESCENT IN UNILATERAL TRANS-TIBIAL AMPUTEES

7.1 Introduction

Currently, the majority (~85%) of clinically available prosthetic feet are rigidly attached to the pylon, however, for more active amputees' prosthetists could prescribe articulated ankle-foot attachments. The attainment of foot-flat after initial contact (IC) prosthetic foot simulated 'plantar-flexion' only by deformation of the heel (compressing material or a heel-keel spring) but also in articulated ankle-foot attachments as a result of the mechanism of the 'ankle'. Previous studies have reported that prosthetic feet with an elastically articulating ankle-foot provides lower-limb amputees certain biomechanical benefits to overground gait compared to using a rigid ankle-foot (Nolan and Lees 2000; Zmitrewicz et al. 2006). Recently, investigators have examined that dynamic response, hydraulically or elastically articulated feet that provide biomechanical benefits compared to habitual prosthetic feet. Use of these devices in unilateral trans-tibial amputees (TT) has led to increases in walking speed and toe clearance, reducing the compensatory mechanism of intact-limb kinetics and the braking effect exerted by the prosthetic limb for overground gait (Portnoy et al. 2012; De Asha et al. 2013a; De Asha et al. 2013b; De Asha et al. 2014; Johnson et al. 2014). However, there is a drawback to these types of passive prosthetic feet, the resistance of 'plantar-flexion and 'dorsi-flexion' simulations are set-up according to overground level gait with the self-selected walking speed of the amputee (Vrieling et al. 2008). Ramp descent locomotive pattern is distinct from overground gait (Lay et al. 2006) and includes an increase of loading response knee flexion that places greater demands on the knee extensors to contract eccentrically (Lay et al. 2007; Vickers et al. 2008). This drawback of prosthetic feet made ramp descent more demanding for TTs as this muscle group was affected by amputation (Winter and Sienko 1988). The amount of residual-knee flexion needed to achieve foot-flat can be reduced by using a foot with a more compliant heel which increases the braking force at the early stance phase during overground gait (Silverman and Neptune 2012; Fey et al. 2013). The increased braking force at the heel rotates the shank forward, with reduced forward velocity of the body's centre-of-mass (CoM) from initial contact to mid-stance stance (De Asha et al. 2014). The research of Su and colleagues reported that lower-limb amputees with a 'rubber-snubber' articulation attachment in the ankle-foot device have easier slope descent (Su et al. 2010).

During slope descent, the prosthetic heel deformation is insufficient to attain foot-flat quicker, so the knee flex and ‘pushed’ forward (Vickers et al. 2008; Fradet et al. 2010). This potentially leads to knee instability within the increased load on the residuum (Perry et al. 1997; Vickers et al. 2008). This is likely to explain why for TTs slope descent is a more demanding task than slope ascent (Macfarlane et al. 1991; Sin et al. 2001; Vickers et al. 2008; Vrieling et al. 2008). As a result, the objective for developers’ of lower-limb prosthetic devices was to create a prosthetic foot that could adapt to lower-limb amputees gait pattern and environment. The functionality of this prosthesis in slope descent has to be a compromise between minimising energy consumption and maintaining stability (Hunter et al. 2010) as lower-limb amputees have to adapt their gait pattern to correspond to their prosthesis functionality and environment.

The microprocessor-controlled quasi-passive hydraulic dampening ankle-foot prosthetic device (*Elan*; Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) was designed to be able to adapt during the stance phase to inclined surfaces Chapter 3.12. Therefore, use of the adaptable device compared to non-adaptable should improve the biomechanics of ramp descent in TTs. Previous studies of the adaptive ankle-foot device ‘Proprio-Foot’ (Ossur hf, Reykjavik, Iceland) investigated its benefits during ramp descent (Fradet et al. 2010; Darter and Wilken 2014; Agrawal et al. 2015). However, the adaptation of ‘Proprio-Foot’ prosthetic device occurs only during the swing phase, but during the stance phase acts as a conventional dynamic-response foot (Au et al. 2007b; Versluys et al. 2008; Eilenberg et al. 2010). Amputee participants in the study of Fradet and colleagues have reported ‘feeling safer’, but biomechanical results indicated negligible changes on lower-limb joints (Fradet et al. 2010). Other studies have termed, that participants with ‘Proprio-Foot’ have enhanced symmetry between limbs and reduced the energy cost (Darter and Wilken 2014; Agrawal et al. 2015). Nevertheless, there are no investigations of a prosthetic ankle-foot that has adapted to terrains during the stance phase.

The purpose of the study was to determine how *Elan* (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) hydraulic dampening ankle-foot device in active (MC-AF) mode compared to non-active (nonMC-AF) mode and to elastically articulated ankle-foot device (elastic-AF) (*Epirus;* Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) would affect the biomechanics of TTs during the ramp descent. Where the *Elan* device in non-active mode (nonMC-AF) performs as an *Echelon* device (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). The characteristics of the assessed ankle-foot prosthetic devices are described in the general methodology section (Chapter 3.12). This study tested the hypothesis, if attainment of foot-flat quicker, the reduction pylon/shank forward rotation with the reduction of knee flexion (Chapter six) during ramp descent would reduce mechanical power at the knee with MC-AF ankle-foot articulation compared to non-adaptive (elastic-AF or nonMC-AF) ankle-foot articulations. Another question in this research was, if reduced pylon/shank forward rotation would also reduce the impact on the intact side for MC-AF compared to non-adaptive (elastic-AF, nonMC-AF) ankle-foot articulations. Because pylon/shank controlled forward rotation on the prosthetic side would influence landing on the intact foot.

7.2 Methods

7.2.1 Participants and Ethics

In the current study, nine active TTs males participated (mean (SD): age - 41.2 (12.9) years, mass - 74.14 (15.7) kg, height - 1.76 (0.06) m) were classed as at least K3 on the Medicare scale. All participants had experience of using articulating ankle-foot prosthetic devices. Details of all participants in the study are presented in chapter 3.4. Ethical approval for this study was conducted in accordance with the tenets of the Declaration of Helsinki and granted from the University of Bradford's Committee for Ethics in Research.

7.2.2 Specific equipment, procedure, data acquisition and processing

All amputee participants were familiarised with each prosthetic device (*Epirus, Elan*) described in chapter 3.12. Details of each prosthetic device (*Epirus, Elan*) with full walking protocol are presented in chapter 3.12. Participants completed six successful trials for each condition, and for each limb (prosthetic and intact). There were two blocks completed (*Epirus* and *Elan*) in counterbalanced order across participants. Within the block, the first had always been self-selected walking speed as they would normally walk down the ramp and the second comfortable slow walking speed. The block with the *Elan* was collected when the microprocessor was active (MC-AF) or inactive (nonMC-AF) modes which were done in random order and ‘blindly’ for participants via Bluetooth connection. The walking protocol with full specifications of *Epirus* (elastic-AF), *Elan* in active (MC-AF) and non-active (nonMC-AF) modes is presented in chapter 3.12.

To examine ramp descent, a custom made modular ramp 2.8 metres long, with an inclination of 5 degrees and 1.0-metre long level ground landing was used. To validate ‘force structure’ within the acceptable boundaries CalTester technique was used (C-Motion, Germantown, MD, USA) (Chapter 3.7.3). The full ramp specification is presented in chapter 3.7.6.

Kinematic and GRF data were recorded at 200 Hz and 400 Hz respectively. Vicon Nexus 1.8 software was used to record and label data (Vicon, Oxford, UK). After labelling C3D files transferred to Visual 3D software (C-Motion, Germantown, MD, USA) where nine segment 6DoF (six degrees of freedom) (Cappozzo et al. 1995) was created, and further data extractions were performed (Chapter 3.13). A functional joint centre approach was employed to calculate anatomical joint centres (Schwartz and Rozumalski 2005). The distal end of pylon was created at the same height as the functional joint centre of the intact ankle. Unified deformable segment' (*UDS*) method was used to calculate the prosthetic 'ankle' power without FJC evaluation at the 'ankle joint' (Takahashi et al. 2012; Takahashi and Stanhope 2013). However, the method allows the calculation only of scalar power during the stance phase. The shank/pylon was assumed to be rigid, and everything distal to the shank/pylon is deformable, so the calculation of power does not depend on the type of prosthetic foot attachment. The distal end of the pylon is the physical application point of the forces and moments transferred to and from the shank/pylon. Prior evaluation of prosthesis foot powers was applied the UDS method to the pipeline of Visual 3D software (C-Motion, Germantown, MD, USA). To calculate UDS power the command 'Compute_UD_Power' was used in the software pipeline. The software provides a method to automate via the pipeline tool, where a pipeline is a command language. More detailed description of the data recording and processing is presented in chapter 3.13.

To filter kinematic and kinetic data a 4th order zero-lag Butterworth low-pass filter cut-off frequency of 6 Hz and 20 Hz was used accordingly. To define the stance phase vertical components (Z) of ground reaction forces were used with a threshold of 20 N from initial contact (IC) till toe-off (TO). Kinematic data were used to define single-limb-support (SLS) (due to only one available force plate) from contralateral foot TO till the IC. There contralateral TO was created according to the Zeni gait event detection algorithm based on toe marker velocity relative to the pelvis (Zeni Jr et al. 2008), and IC was created at the threshold of the contralateral heel virtual marker's vertical velocity reduced below 0.15 m/s. The full protocol procedure used laboratory equipment and the recorded data processing is displayed in chapter three.

7.2.3 Data analysis

The variables extracted for each trial were averaged across the trials to provide the main parameter for each condition per participant. The study examined the residual limb (hip, knee) with three prosthetic articulations (elastic-AF, nonMC-AF, MC-AF). The following variables were calculated for the prosthetic side during a stance phase. Mechanical work at the biological joints (hip and knee) was the integral of their sagittal plane negative or positive joint power. The work at the prosthetic ‘ankle’ UDS was defined as power integrals of its negative or positive scalar power. The following variables were calculated for the intact side during a stance phase. 1st and 2nd vertical peaks (Fz). Peak braking (Fy) and peak propulsion (Fy). All work and GRFs data were normalised to the participant’s body weight.

7.2.4 Statistics

The data were tested for normal distribution using the Kolmogorov-Smirnov test and shows that the distribution of the data is normal. To investigate if participants’ having familiarity with habitual ankle-foot devices (*Elan* or *Echelon VT*) as a ‘between factor’ in a mixed-design of repeated measures analysis of variance (ANOVA) have an effect on results. There were four participants in each group. The participant with a Re-flex Rotate habitual foot was included in the *Echelon VT* group. The mean values were determined for each participant and each condition. The mean values were analysed between ankle-foot articulations (elastic-AF, nonMC-AF, MC-AF) and gait modes (comfortable slow and self-selected) as repeated factors in a mixed design ANOVA. Effect size differences (low $d < 0.3$, moderate $0.3 < d < 0.5$ and high $d < 0.5$) were calculated as Cohen’s (Cohen 1988). The analysis was performed in Statistica (v6, StatSoft, Inc., Tulsa, OK, USA). To identify the significance between conditions when the main effects were significant a Tukey honestly significant difference (HSD) post hoc test was used. The level of significance was set at $p < 0.05$.

7.3 Results

7.3.1 Residual limb

Residual limb measurements results are summarised in table 12 below, all statistically significant differences are highlighted in bold. To analyse participants familiarity with habitual ankle-foot design (*Elan* or *Echelon VT*) a 'between factor' in a mixed-design ANOVA was used where ankle-foot articulations and gait modes were used as repeated factors. The analysis indicated that for all except one of the parameters there was no significant main or interaction effect of the habitual foot type for any of the parameters investigated (all $p \geq 0.13$). The exception was a hip negative work that indicated a significant group by ankle type interaction ($p = 0.008$). The interaction indicated: a hip negative work had no difference between groups for the MC-AF and nonMC-AF 'ankle' conditions, but an increase for the habitual *Elan* users and decrease for the habitual *Echelon VT* users in the elastic-AF 'ankle' condition. Therefore, this analysis indicates that habitual ankle-foot design had minimal effect on the presented results so that further analysis will exclude examination of participants' familiarity with habitual ankle-foot.

Prosthetic side negative work at the *UDS* indicated no significant difference between speeds ($F_{(1,8)}$ =0.04, $p = 0.84$) but was significant between ankle articulation types ($F_{(2,16)}$=14.22, $p < 0.001$) (Table 13). There was more negative work done at the *UDS* when using the MC-AF than either nonMC-AF or elastic-AF articulations, and more when using the nonMC-AF compared to the elastic-AF articulation (Figure 36). Knee negative work did not indicate a significant difference between speeds ($F_{(1,8)}$=0.64, $p = 0.45$), but negative work has a trend ($F_{(2,16)}$=2.92, $p = 0.083$) to be reduced at both speeds when using the MC-AF compared to the nonMC-AF or elastic-AF articulation (Figure 36). Hip negative work was significantly lower at slow speed ($F_{(1,8)}$=6.33, $p = 0.036$), but there was no difference between ankle articulation types ($F_{(2,16)}$=0.32, $p = 0.73$) (Table 13).

Table 13. Group mean (±SD) residual limb negative work during single-limb-support phase at the distal end of the prosthetic shank ('ankle') and knee and hip when using MC-AF, nonMC-AF and elastic-AF at slow and self-selected walking speed (SSWS). Where differences between MC-AF and nonMC-AF or elastic-AF articulations are effect sizes Cohen's (d) presented (in *italics*).

	Slow			***SSWS***			p *value (F value)*
	MC-AF	*nonMC-AF*	*elastic-AF*	*MC-AF*	*nonMC-AF*	*elastic-AF*	
UDS (J.kg^{-1})	-0.158 (0.085)	-0.114 (0.041) *0.6*	-0.088 (0.039) *1.0*	-0.143 (0.052)	-0.125 (0.055) *0.3*	-0.088 (0.017) *1.2*	Speed 0.84 (0.04) **Pros < 0.001 (14.21)** Int. 0.29 (1.35)
knee (J.kg^{-1})	-0.004 (0.004)	-0.010 (0.012) *0.7*	-0.019 (0.028) *0.7*	-0.011 (0.012)	-0.019 (0.018) *0.5*	-0.017 (0.011) *0.5*	Speed 0.45 (0.64) Pros. 0.083 (2.92) Int. 0.33 (1.19)
hip (J.kg^{-1})	-0.010 (0.005)	-0.015 (0.011) *0.6*	-0.018 (0.009) *0.9*	-0.026 (0.022)	-0.022 (0.021) *0.2*	-0.026 (0.014) *<0.1*	**Speed 0.036 (6.33)** Pros. 0.73 (0.32) Int. 0.37 (1.05)

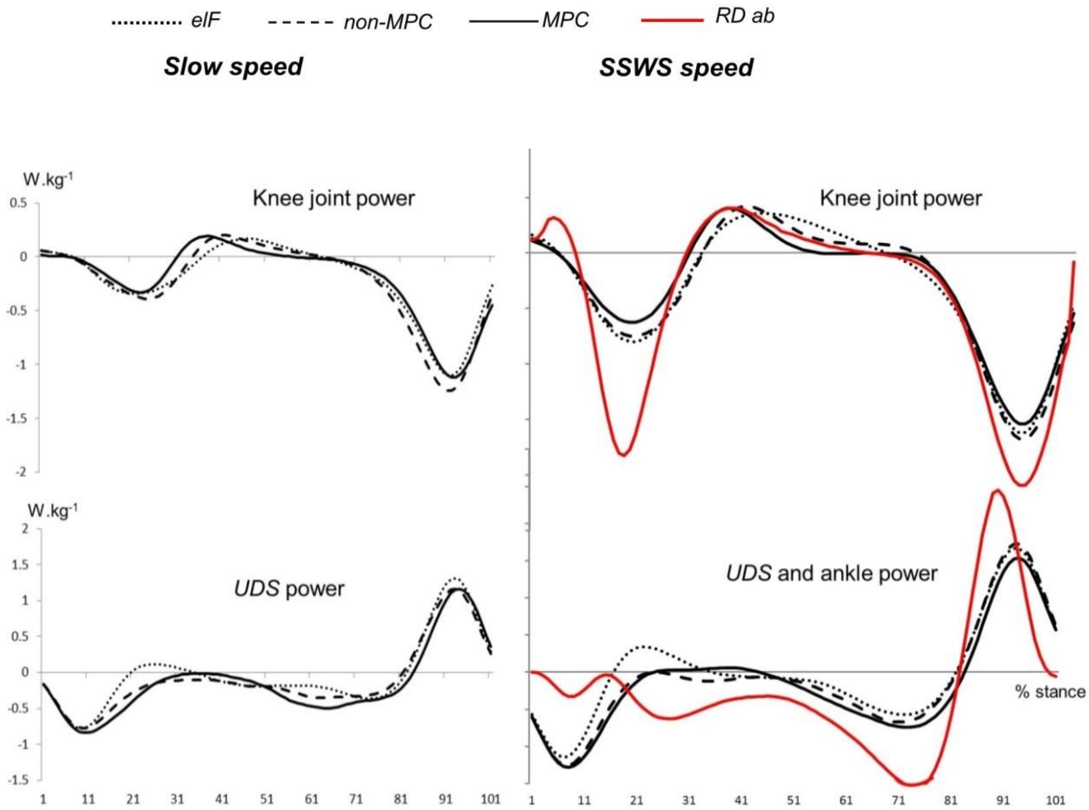

Figure 36. Ensemble group mean stance knee joint power and UDS power when using the elastic-AF (black dotted line), nonMC-AF (black dashed line), MC-AF (solid black line) ankle-foot and RD ab -ramp descent able-bodied individuals (solid red line). Able-bodied data were obtained from chapter five.

7.3.2 Intact limb

Intact limb measurements are summarised in table 13 (below), all statistically significant differences are highlighted in bold. Cohen's effect size (d) presented (in italics). GRF profiles of intact side plotted in Appendix 11.

There were no main effects of prosthetic articulation in GRF values (normalised to body weight): 1st vertical peak, 2nd vertical peak, peak braking or peak propulsion. There were no interactions between prosthetic articulations and speed. Cohen's effect size (d) low between MC-AF and nonMC-AF or elastic-AF articulations.

Table 14. Group mean (±SD) intact limb, GRF values (N/kg) normalised to body weight: 1st vertical peak (Fz), 2nd vertical peak (Fz), peak braking (Fy), peak propulsion (Fy), when using MC-AF, nonMC-AF and elastic-AF at slow and self-selected walking speed (SSWS). Where differences between MC-AF and nonMC-AF or elastic-AF articulations are effect sizes Cohen's (d) presented (in *italics*).

	Slow			***SSWS***			p *value* (*F value*)
	MC-AF	*nonMC-AF*	*elastic-AF*	*MC-AF*	*nonMC-AF*	*elastic-AF*	
1st vertical peak (Fz)	1.16 (0.11)	1.17 (0.13) *0.1*	1.16 (0.12) *0.1*	1.36 (0.17)	1.33 (0.14) *0.2*	1.34 (0.14) *0.2*	**Speed<0.001 (35.34)** Pros. 0.34 (1.16) Int. 0.43 (0.88)
2nd vertical peak (Fz)	1.03 (0.04)	1.02 (0.05) *0.1*	1.02 (0.06) *0.1*	1.08 (0.08)	1.07 (0.08) *0.1*	1.07 (0.08) *0.1*	**Speed 0.02 (8.16)** Pros. 0.59 (0.54) Int. 0.98 (0.02)
Peak braking (Fy)	-0.13 (0.04)	-0.13 (0.05) *0.1*	-0.14 (0.03) *0.3*	-0.21 (0.04)	-0.20 (0.04) *0.2*	-0.21 (0.04) *0.1*	**Speed <0.001 (54.81)** Pros. 0.33 (1.18) Int. 0.46 (0.81)
Peak propulsion (Fy)	0.20 (0.05)	0.20 (0.05) *<0.1*	0.20 (0.04) *<0.1*	0.26 (0.05)	0.26 (0.05) *0.1*	0.25 (0.04) *0.2*	**Speed<0.001 (70.03)** Pros. 0.49 (0.75) Int. 0.21 (1.73)

7.4 Discussion

The study determined how adaptive articulated MC-AF compared to non-adaptive ankle-foot articulations would affect the biomechanics of TTs during the downslope gait. Returning to the findings posed in the previous study, it is now possible to state that during ramp descent foot-flat is lowered to the floor following initial contact quickest for elastic-AF ankle-foot then MC-AF ankle-foot articulation. The following single-limb-support phase has a forward shank/pylon rotation that transferring the body weight over the foot indicated that adaptive, articulated MC-AF ankle-foot was slowest compared to nonMC-AF or elastic-AF ankle-foot articulations. This was likely to be caused by increased negative work done at the *UDS* can be drawn from the present study. In addition, knee loading response and negative mechanical power were reduced in early stance phase with adaptive (MC-AF) compared to the non-adaptive (nonMC-AF or elastic-AF) ankle-foot articulations. However, there were no effects of ankle articulation on GRFs of intact side. This combination of findings provides some support for the conceptual premise that is presented below.

To establish foot-flat quicker during ramp descent is crucial to allow prompt ‘ankle’ articulation towards plantar-flexion (Chapter six). Thus, requirements are important for anterior-posterior stability on inclined surfaces which insufficiency could lead to the risk of falling. The MC-AF articulation for down the ramp gait has established foot-flat quicker than nonMC-AF by reduction of ‘plantar-flexion’ resistance which provides a stable base of support for further body weight transition and seems reduces knee peak loading response (Chapter six). Nevertheless, the foot has to stay flat on the ground longer to deliver a stable base of support within dynamic stability. The foot stays flat on the ground (relative to the percentage of stance) longer with the MC-AF compared to the elastic-AF or nonMC-AF devices at slow speed (Chapter six). There maintenance of foot-flat on the ground contributes to dynamic stability which importance increases on inclined surfaces (Vickers et al. 2008).

Gait down the ramp in TTs with a rigid ‘ankle’ prosthetic device delivers the sensation of ‘pulling’ the residual knee forward. The knee ‘pulls’ forward as compensation to attain foot-flat quicker (Vickers et al. 2008; Vrieling et al. 2008). Certainly, amputees with an articulated ‘ankle’ would attain foot-flat quicker, which would be a result not only of the heel deformity but also articulation of the ‘ankle’. Earlier research presented that TTs have reported it was ‘easier’ to approach down the ramp with an articulated ‘ankle’ attachment compared to non-articulated (Su et al. 2010) and ‘safer’ with a prosthetic foot that can ‘plantar-flex’ during a swing phase (Fradet et al. 2010). Attainment of foot-flat is crucial during down the ramp gait for TT’s; the study findings show that time to attain foot-flat was fastest when using the elastic-AF articulation and slowest when using the nonMC-AF articulation. Although, the foot-flat was attained faster when using the MC-AF compared to the nonMC-AF articulation. A number of studies have already demonstrated the benefits of hydraulic ‘ankle’ in overground gait (De Asha et al. 2013a; De Asha et al. 2013b; Johnson et al. 2014). The finding of the present study suggests that the MC-AF ankle-foot articulation (ramp descent adaptive mode) has reduced hydraulic damping in ramp descent compared to conventional hydraulic nonMC-AF ‘ankle’. Hence, the MC-AF (adaptive mode) ankle-foot articulation can ‘plantar-flex’ the prosthetic foot quicker than a conventional hydraulic ‘ankle’ which was set for overground gait. In the MC-AF ’ankle’ with active ramp descent mode after the attainment of foot-flat followed the increase of ‘dorsi-flexion’ hydraulic dampening that control’s shank forward rotation within body weight transfer over the support prosthetic foot. With the elastic-AF articulation attainment of foot-flat is performed through a combined mechanism, deformation of heel spring and ‘rubber-snubber’ ball hinge to ‘plantar-flex’ and attain foot-flat. However, after attainment of foot-flat, the ‘rubber-snubber’ ball joint would ‘pull’ the shank/pylon forward to the neutral position as a result of the elastic recoil properties of the ‘rubber-snubber’ hinge material. The neutral position is relative to the foot and pre-set by the prosthetist. The ‘rubber-snubber’ hinge recoils the shank/pylon motion towards ‘dorsi-flexion’; this is restricted by ‘hard stop’ so ‘dorsi-flexion’ in an elastic-AF prosthetic device would be due to deformation of the fore-foot keel.

The current study found that the amputees with the adaptive MC-AF compared to non-adaptive (nonMC-AF and elastic-AF) articulation have reduced residual-knee loading response flexion (Chapter six). A possible explanation for these results is likely to be a combination of the MC-AF function: reduced resistance of the foot to ‘plantar-flex’ that followed by increased ‘dorsi-flexion’ resistance. The MC-AF compared to nonMC-AF, or elastic-AF articulations have increased the ‘plantar-flexion’ resistance during single-limb-support which was verified by a reduction of shank/pylon angular velocity (Chapter six). The results of the present study draw attention to the significant increase of negative work done during single-limb-support by the adaptive MC-AF compared to non-adaptive (nonMC-AF or elastic-AF) and greater nonMC-AF compared to the elastic-AF articulation. The evidence presented thus far supports the idea that 'plantar-flexion' and knee extension coupled are likely to be affected by the increase of negative work at the 'ankle'. The 'ankle' negative work was greater with the MC-AF compared to the nonMC-AF or elastic-AF and greater to the MC-AF and nonMC-AF compared to elastic-AF articulation. The most likely explanation of this result is a reduction of knee negative work (trend, ($F_{(2,16)}$=2.92, p = 0.083)) with the MC-AF 'ankle' compared to the nonMC-AF. The participants with the MC-AF 'ankle' attachment have reduced compensation at the knee as a result of the prosthetic 'ankle' contribution. Hence, the MP-AF is more appropriate for ramp descent gait than other non-adaptive 'ankles' as it is likely to improve dynamic stability during the stance phase. Certainly, *UDS* power should not have quantities comparable to able-bodied ankle powers. However, what is interesting that in Figure 36 it indicates that for able-bodied the ankle provides minor negative work in early stance phase, but increases in mid-stance which is distinct to the *UDS* ankle-foot. The research of De Asha et al. (2013) has shown that the hydraulically articulated 'ankle' prosthetic device improves walking speed and as a result efficiency of gait without improving the symmetry of gait pattern (De Asha et al. 2013b). Hence, there is a similar correlation between a prosthetic 'ankle' and able-bodied ankle that could likely be related to ramp descent.

Prosthetic devices with rigid or elastic 'ankle' during overground gait typically deliver a small burst of positive power in early stance from heel-keel recoil (De Asha et al. 2013b). However, this positive burst from the heel-keel could not be used for propulsion as it occurs too early to contribute to propulsion. Also, this burst could lead to premature heel rise, which is common with TTs. The use of the hydraulic ankle-foot device was shown to reduce this burst (De Asha et al. 2013b) as the recoil of heel-keel energy dissipated within the hydraulic dampening mechanism of the 'ankle'. This burst in early stance could have additional negative effects in ramp descent increasing shank/pylon forward rotation and as a result knee flexion. From Figure 36 of *UDS*, power graphs it can be observed that the elastic-AF articulation provided positive work in early stance unlike nonMC-AF and MC-AF hydraulic articulations of the *Elan* device. The differences between hydraulic and elastic articulations are highlighted in Figure 36 and indicate the benefits of the hydraulic 'ankle' over elastic articulation.

GRFs are crucial for inverse dynamics calculations (Gordon et al. 2004). Although GRFs were not measured for the affected side in the current study, previous research showed that GRFs on the intact up to 23% greater than on prosthetic side (Gailey et al. 2008) which increased on the ramp descent compared to overground gait (McIntosh et al. 2006; Lay et al. 2007; Franz et al. 2012). Agrawal et al. (2015) presented in the study, the GRFs of intact side show compensation mechanism which depends on prosthetic foot functionality (Agrawal et al. 2015). This compensation could lead to overload on an intact limb and as a result cause pain and predispose the patient to premature degenerative diseases, such as osteoarthritis (Hurwitz et al. 2001). Contrary to expectations, this study did not find a significant difference between ankle articulations on the intact limb ($p<0.34$), but the clinical significance of GRFs has to be interpreted with caution (Menard et al. 1992). There are a few possible explanations for the no effect of prosthetic articulation type on the intact limb. Recruited amputee participants were from the experienced Alpha and Beta Testing group and possibly able to maintain better between limb symmetry. Interestingly, according to Lehmann (1993), dynamic response prosthetic feet tend to reduce the first vertical peak of GRFs on intact side, and so reduce the impact on an intact limb (Lehmann et al. 1993). However, the research examined SACH versus dynamic response feet, but this study examined the difference between prosthetic ankle-foot articulation rather different types of 'heel' absorption. Hence, examination effects of ankle-foot articulations on GRFs of the intact side could be offered as fine outcomes, but the number and ability of participated amputee patients could have an effect on this outcome. Furthermore, the compensatory mechanism could be presented in individual joints (ankle, knee and hip) of contralateral limb (Nolan et al. 2003), which kinetic could present additional information about effects of the ankle-foot prosthesis on contralateral limb.

The current research was not specifically designed to evaluate the difference between ramp descent and overground gait. However, Figure 36 included data (solid red lines) of able-bodied individuals (Chapter five) during ramp descent with self-selected walking speed. Data were added to differentiate between TT and able-bodied individuals. Overground gait was excluded from this study design, but important effects of prosthetic devices on gait are described below. Early studies clearly demonstrated that the increase of the knee flexion for ramp descent compared to overground gait in TT is performed to attain foot-flat quicker (Vickers et al. 2008; Vrieling et al. 2008; Fradet et al. 2010). Thus knee flexion during the stance phase could potentially lead to condensed volume compared to the intact side (Isakov et al. 2000). The study of overground gait indicated that TTs have a longer weight bearing phase which could explain longer EMG activity in co-contraction of the residual-limb hamstrings and quadriceps to stabilise the knee with increased energy expenditure (Isakov et al. 1996a; Isakov et al. 2000). This co-contraction is increased during downslope gait (Vickers et al. 2008). Co-contraction has an insignificant involvement in forward propulsion and depends on the prosthetic foot functionality (Barth et al. 1992). Therefore, it could be that an increase of co-contraction during ramp descent in TT depends on the prosthetic design. Thus, the increase of co-contraction activity potentially leads to an increase in energy expenditure. The results of this study show that amputees with the MC-AF compared to nonMC-AF or elastic-AF articulation have reduced residual knee flexion and negative work (trend, $p = 0.083$). Hence, the MC-AF compared to non-adaptive (nonMC-AF and elastic-AF) articulation users could potentially reduce energy cost when walking down a ramp. It also has to be recognised that TTs have reduced negative work (Figure 36) compared to able-bodied individuals (Chapter five) this was probably the result of partly amputated flexor muscles (Winter and Sienko 1988). The MC-AF functioned to preserve energy expenditure, and as a result, controlled contribution at the residual-knee. It is also important to note that the MC-AF device (~ 1.2 kg) is approximately 0.3 kg heavier than nonMC-AF and 0.8 kg heavier than the elastic-AF prosthetic device. The weight of the prosthetic could have an effect on energy cost during a swing phase, as heavier prosthetic devices assist the propulsion of the trunk forward (Gitter et al. 1997; Lehmann et al. 1998) without notably increasing metabolic cost (Gailey et al. 1997) in overground gait. A possible benefit of a heavier prosthesis could include maintenance of balance during amputees' locomotion. Heavier prosthesis with improved functionality has shown a reduction of metabolic cost of prosthetic devices for transfemoral amputees (Buckley et al. 1997) and TT (Au et al. 2009) in overground gait. However, to confirm this future work is required.

7.5 Conclusion

The microprocessor controlled hydraulic ankle-foot device during ramp descent delivered greater negative work at the 'ankle' compared to non-adaptive hydraulic and elastic ankle-foot articulation types. This was a combination of reduced resistance towards 'plantar-flexion' to attain foot-flat quicker, which is then followed by increased resistance towards 'dorsi-flexion' to reduce shank/pylon angular velocity forward. There was then a corresponding reduction in flexion and negative work at the residual-knee which reduces biomechanical compensations during ramp descent in TTs. The reduced compensation improves the dynamic stability and potentially reduces energy cost during ramp descent in TT's with such prosthesis. The intact side GRFs did not present effect between ankle-foot articulations during ramp descent. This suggests that active amputees with MC-AF articulation type may have functional advantages during the stance phase on a prosthetic limb in ramp descent. The study findings have also validated the use of a unified deformable segment (*UDS*) model in ramp descent gait which should be valuable to quantify comparative performance between different designs of prosthetic 'ankle' components.

CHAPTER EIGHT - FINAL DISCUSSION

8.1 Final discussion

The main experimental focus introduced to the reader in the thesis is sixth and seventh chapters. These were designed to investigate the effects of a microprocessor-controlled, hydraulically damped, uniaxial articulating prosthetic ankle-foot *Elan* (MC-AF) device in active TTs. The *Elan* device articulation was compared to conventional (non-adaptive) hydraulically (nonMC-AF) and elastically (elastic-AF) articulations on the biomechanics of ramp descent. The main findings of this thesis are that the use of the MC-AF articulation compared to nonMC-AF or elastic-AF articulations have greater negative work at the 'ankle' which reduces residual-knee loading response flexion, negative mechanical power in early stance and minimum knee flexion during single-limb-support. The findings also demonstrated that the Centre-of-Pressure forward velocity beneath the prosthetic device during single-limb-support showed improved body motion control and smoother transition with MC-AF compared to nonMC-AF or elastic-AF. Surprisingly, no differences were found in symmetry between limbs or GRFs of a contralateral limb between ankle-foot articulations. Hence, the use of MC-AF presented advantages only on the residual limb in this study. The introductory experimental fourth and fifth chapters were designed to investigate biomechanics between overground and ramp gait in healthy, able-bodied individuals with the unilateral restricted ankle to simulate unilateral TTs. The findings were presented, that 'ankle' contribution did not have an effect on VL during single-limb-support. To compensate inapt 'ankle' function, knee flexion loading response was increased as a result of the foot-flat delay on the ground, which also leads to an increase of positive knee work during single-limb-support as a contraction. The overall aim of the study was to gain insight into the prosthetic ankle-foot articulations during ramp descent by synthesising analysis of lower-limb biomechanics and whole body motion. These findings and their relevance are discussed below.

The part of this study set up to assess the importance of slow ramp descent for TTs. When TT participants descended the ramp at comfortable slow walking speed, time to foot-flat was longer compared to self-selected walking speed.

The time to foot-flat in ankle-foot devices were a result of heel keel deformation, and ‘ankle’ mechanism articulation (i.e., simulated ‘ankle’). The insufficiency of simulated ‘plantar-flexion’ would require compensatory knee-flexion which would likely affect dynamic stability (Perry et al. 1992). Thus, rapid/smooth attainment of foot-flat is critical for safe ramp descent (Perry et al. 1997; Sin et al. 2001). The increased stance time on the prosthetic side during slow compared to self-selected walking speed ramp descent (Chapter six) also could be problematic for TTs, because TTs tend to reduce load on prosthetic compared to intact limb in overground gait (Murray et al. 1983; Engsberg et al. 1993; Silver-Thorn et al. 1996; Nolan et al. 2003). However, the use of the hydraulically articulated ankle-foot device has delivered more advantageous benefits because time-dependent hydraulic articulation reduces socket pressure (Portnoy et al. 2012) for transfer body weight onto the prosthetic limb in a smoother, less faltering manner (De Asha et al. 2013b). Previous research also indicated that a reduction of walking speed increases stance time on an intact limb when compared to the prosthetic limb in overground gait so as a result asymmetry between limb increases (Nolan et al. 2003). To maintain the residual knee in a flexed position, TTs employ co-contraction of the hamstrings and quadriceps to stabilise these knee (Isakov et al. 1996a; Isakov et al. 2000). This co-contraction controls knee flexion in order to provide controlled weight acceptance on the prosthetic limb, so as result TTs have a higher metabolic cost of ambulation compared to able-bodied individuals (Barth et al. 1992). The amplitude of the associated muscle EMG activity becomes increased during ramp descent and increased further with slower, more controlled walking speed (Vickers et al. 2008). These suggestions seem to be consistent with other research which found that reduced knee flexion during weight acceptance leads to reduced energy expenditure in overground gait (Waters and Lunsford 1985). It is possible to suggest that, slow ramp descent requires more control to transfer body weight over the support limb due to the prolonged control of the potential gravitational energy (Chapman 2008) compared to overground gait or ramp descent with self-selected walking speed. The slower attainment of foot-flat, increased stance time and increased knee flexion highlights that slow ramp descent was a more demanding task for TTs compared to self-selected walking speed.

Contrary to expectations, the examination of symmetry between limbs (Chapter six) did not find a significant difference (two-way ANOVA or effect size) between prosthetic ankle-foot articulation types on stance time, step length, knee flexion or CoM forward velocity during stance. The findings of chapter four are consistent with chapter six, which also did not present the effect of ‘ankle’ articulation on those parameters. For this reason, it is possible that prosthetic ankle-foot articulation might not have a significant effect on the spatio-temporal parameters of contralateral side and as a result of the symmetry between limbs during ramp descent. Prosthetic devices were utilised the same heel and fore-foot keels within a participant with ‘ankle’ articulation mechanisms even if those mechanisms have different types of articulations. Controversially, the study of Agrawal et al. (2015) suggested that the functionality of prosthetic feet has an effect on symmetry between limbs during ramp descent (Agrawal et al. 2015). However, that study compared prosthetic feet with rigid ‘ankle’ versus adaptable ‘Proprio-Foot’ (Ossur hf, Iceland), so did not compare between prosthetic ‘ankles’ but prosthetic feet (advanced (adaptable) versus rigid (not articulated) prosthetic foot devices). Further examination also confirms that ankle-foot articulation types did not have a significant effect on GRFs of the intact limb (Chapter seven) during ramp descent. The most likely explanation is that prosthetic devices were utilised the same heel and fore-foot keels within a participant and have articulated ankle-foot mechanism even if the mechanism used different types of articulations. The results are similar to those represented in chapter five. Here additional evidence that suggested that restricting the ankle does not have an effect on contralateral limb work. The only exception that may exist is that with a restricted ankle, the contralateral hip reduces work for a period of 3rd rocker ($p = 0.01$), however, a low effect size ($d \leq 0.2$) does not support it. The present findings also seem to be consistent with other research which found that TTs with different prosthetic foot devices had no significant effect on EMG activity for the intact limb, but there were effects on the residual limb according to the used prosthetic foot (Barth et al. 1992). Conversely, the findings of the current study do not support the previous researchers, where they have presented the effects of prosthetic ankle function on the intact limb (Vickers et al. 2008; Agrawal et al. 2015). However, those studies examined rigid ‘ankle’ (SACH) and single axis ‘ankle’ (SA) (Vickers et al. 2008) or rigid ‘ankle’ versus adaptable ‘Proprio-Foot’ (Ossur hf, Iceland) (Agrawal et al. 2015), so there was not an assessment of prosthetic ‘ankle’ functionality. Lehmann et al. (1993) have presented that dynamic response prosthetic feet tend to reduce the first vertical peak of GRFs on intact side, and so reduce the impact on the intact limb (Lehmann et al. 1993). However, again the research examined SACH versus dynamic response feet, but this study examined different types of ‘heel’ absorptions rather differences between prosthetic ankle-foot articulations. Nevertheless, ramp descent has greater demand on lower-limbs than overground gait (McIntosh et al. 2006; Lay et al. 2007; Franz et al. 2012). There potential overload of the intact side could lead to asymmetry in musculoskeletal function and cause pain and predispose patients to premature degenerative diseases, such as osteoarthritis (Hurwitz et al. 2001). To compensate inapt ankle-foot articulation could be employed individual joints (ankle, knee and hip) of contralateral limb (Nolan et al. 2003). Detailed examination of the joint kinetics of intact-limb during ramp descent may identify those compensations.

The obtained data suggested that the objective of increased limb negative rotational work in the 1st and 2nd rockers during ramp descent is to control body weight transition within increased potential gravitational energy (Chapter five). These results corroborate with the findings of study Kuo and Donelan (2010), who demonstrated that the negative work of the lower-limb is dependent on the amount of CoM vertical displacement in the arc in the inverted pendulum model (Kuo and Donelan 2010). Findings presented (Chapter six), that reduced resistance of articulation at the prosthetic ‘ankle’ after initial contact when using MC-AF would allow attaining foot-flat sooner compared to nonMC-AF articulation. The attainment of foot-flat sooner after initial contact without the additional requirement to compensate at the knee should improve residual knee stability during loading response. The use of the MC-AF ankle-foot articulation reduces compensatory requirements of knee flexion and reduces this effect irrespective of speed compared to the other two attachments (Chapter six). On the other hand, increased knee flexion during single-limb-support could potentially also reduce residual knee stability during stance because of the increased load it places on the residuum (Perry et al., 1997; Vickers et al., 2008). The shank forward rotation over the support limb deliver control individuals to ‘fall’ forwards during single-limb-support and contributes into dynamic stability (Perry et al. 1997).

Loading response knee flexion was reduced for MC-AF compared to nonMC-AF (Chapter six). However, the MC-AF was the second fastest articulation after elastic-AF (*Epirus*). The elastic-AF articulation has ‘rubber-snubber’ properties that allow to attain foot-flat quicker which was an efficient response during the early stance phase on the 5 degrees of inclination ramp. Throughout the single-limb-support, the ‘rubber-snubber’ will tend to recoil, return to the neutral position and so ‘pull’ the shank/pylon forward. The ‘’pull’ of the shank/pylon would lead to an increase in minimum knee flexion when an elastic-AF was used. Where the use of MC-AF articulation after the attainment of foot-flat provides an increase of ‘dorsi-flexion’ resistance so reduces shank/pylon rotation forwards. Chapter six indicates that the mean forwards shank angular velocity during single-limb-support was reduced for MC-AF compared to the other two articulations. This also supported by data from chapter seven, where MC-AF articulation provides an increase in UDS negative work with a reduction in compensations at the knee. Chapter four also presented data, that restricted ankle compared to non-restricted in able-bodied participants have increased loading response knee flexion. Due to the restricted ankle ‘pulls’ the shank forward, which leads to an increase of loading response knee flexion to attain foot-flat sooner. The finding is consistent with findings of past studies by Su et al. (2010), which suggested that ramp descent with an articulated ‘ankle’ is more favourable than with rigid devices for amputees (Su et al. 2010). Vickers et al. (2008) research also support (Vickers et al. 2008) that to attain foot-flat sooner with restricted ankle participants require an increase in loading response knee flexion (Chapter Four). Therefore, not only the attainment of foot-flat quicker but also reduction of the shank/pylon rotation forward could be presented as fundamental parameters for safe and controlled ramp descent.

Detailed examination of VL angular velocity in the single-limb-support during ramp descent did not present influences of ankle-foot articulation types (Chapter six) in TTs or restricted ankle (Chapter four) conditions. A possible explanation for this might be that the knee flexion during ramp descent have controlled forward motion. The consistent findings throughout the experimental chapters indicated that restricted compared to non-restricted ankle condition and nonMC-AF or elastic-AF compared to MC-AF has increased knee loading response flexion alongside the increased knee joint work during single-limb-support. Another explanation for this might be that to maintain whole body motion relative to the supporting foot, knee loading response flexion in conjunction with knee work which was done during single-limb-support compensated for the 'ankle'/ankle functionality (Chapter five and seven) to provide safe and efficient body transition. This suggests the knee flexion compensates according to 'ankle'/ankle functionality to provide control of the Centre-of-Mass motion relative to the support foot (VL). The findings provide support for the conceptual premise that the Centre-of-Mass transition on declining surfaces during single-limb-support is controlled by knee flexion which contribution depends on ankle function. Hence, if in overground gait, the ankle acts as a fulcrum in the inverted pendulum model to minimise energy expenditure (Cavagna et al. 1963; Cavagna and Margaria 1966), but in ramp decent 'ankle/ankle' contribution has changed as the ankle has to provide control to achieve safe body transition with minimal energy expenditure.

The study (Chapter six and seven) findings revealed that use of the MC-AF articulation lessens knee involvement compared to nonMC-AF or elastic-AF, which could likely lead to a reduction of energy cost for MC-AF users. The assessment of energy cost is not part of this thesis. However, the reduction of residual-limb knee joint mechanical work per metre travelled when using the MC-AF (Chapter seven) compared to nonMC-AF or elastic-AF articulations during the ramp descent would suggest that the use of the MC-AF devices could lead to a reduction in energy cost. It can be suggested that use of the MC-AF articulation would reduce metabolic energy expenditure during ramp descent despite the prosthetic foot absorbing more and returning less energy, as controlled ramp descent require more energy absorption than propulsion (McIntosh et al. 2006; Franz et al. 2012). The similar reduction in energy cost was also shown when using the non-restricted compared to restricted ankle condition (Chapter five). There reduction of the knee joint work during single-limb-support is also likely to lead to some reduction in energy cost.

Additionally, reduced the Centre-of-Pressure forward velocity (Chapter six) within increased the power absorption during single-limb-support when utilised the MC-AF compared to nonMC-AF or elastic-AF articulations (Chapter seven) is likely to reduce the knee flexion, which could also suggest a reduction of the muscular compensation. This muscular compensation contributes to the knee stability to retain the whole body transition over the support foot. There reduced knee flexion suggests improvements in dynamic stability. The findings also support that the use of the MC-AF compared to non-adaptive articulations delivers smoother Centre-of-Pressure forward progression during ramp descent (Chapter six). The research of De Asha et al. (2013) also supported that use of hydraulically articulated ankle-foot attachments provide smoother stance phase transition (De Asha et al. 2013a). Chapter six results provide further insight into the research of Winter and Sienko (Winter and Sienko 1988) that the prosthetic 'ankle' component controls the Centre-of-Pressure forward progression (Chapter six) not only in overground gait but also during the ramp descent. Therefore, malfunction of the prosthetic foot may be one of the reasons why lower-limb amputees have higher energy expenditure compared to able-bodied individuals (Barth et al. 1992; Waters and Mulroy 1999; Nolan et al. 2003; Hsu et al. 2006). The research of Darter and Wilken (2014) has shown ramp descent reduces the metabolic energy expenditure when a Proprio-Foot was utilised in active compared to non-active mode (Darter and Wilken 2014). Therefore, the study findings further support the idea of the Centre-of-Pressure forward velocity during single-limb-support delivers a detailed presentation of 'ankle' function.

It seems that results provide support to the claimed benefits of the MC-AF (*Elan*; Chas. A. Blatchford and Sons Ltd, Basingstoke, UK) articulation settings for ramp descent: "*On walking downhill, lower plantar-flexion resistance allows the foot to fully contact the slope sooner for improved safety and security. At the same time, increased dorsiflexion resistance provides a braking effect stabilising the user for a safer, more controlled descent*." (www.endolite.co.uk).

8.2 Limitations and Future directions

The thesis has a number of potential limitations. The current study has only examined sagittal plane gait biomechanics. However, examination of the sagittal plane biomechanics is the main concern in anterior-posterior motion control during ramp descent, and it was supported in the literature. A limitation of the study was that external joint work cannot be directly quantified as the actual work performed by muscles; however, the external work provides direction to the energy cost. The current investigation was limited by the number of participants in experimental chapters six and seven (9 participants). A more thorough study would examine a large, randomly selected sample of unilateral TTs with various causes of amputation. This would improve confidence in the results of this thesis so that the results could be generalised to the wider TT population. Nevertheless, the participant group makes up at least 4% of the annual number of TT traumatic amputee males in the UK (NASDAB, UK, 2011/12). The majority of trans-tibial amputations are due to vascular disease (~85-90%) (National Amputee Statistical Database, NASDAB, UK, 2011/12). Patients who had amputations due to vascular disease are commonly less physically active than traumatic amputees. However, a vascular amputee classified as K3 activity level would be acceptable to participate in the study according to the protocol. This limitation maintained the uniformity of the amputees who participated in the thesis. Traumatic, TT participants were recruited using the criteria that they should be K3-K4 activity level (Chapter 3.12) so could perform the ramp descent tasks independently and without a walking aid. The MC-AF, which was the main focus of the experimental chapters six and seven is prescribed for the lower-limb amputees

K3-K4 activity level. Homogeneity of the participant group may be viewed as a study limitation; however, the group was recruited to avoid confusing results due to different functionality of participants' prosthesis. Certainly, the number of participants in the studies has an effect on the statistical analysis, but a graphical representation of results has supported the applied data analysis. Nevertheless, with a small sample size, caution must be applied, as the findings might not be transferable to the entire lower-limb amputee population.

A potential limitation of the experimental chapters six and seven was the wide range of participant characteristics; including age, weight, residual limb size/length and prosthetic components (stiffness of heel and fore-foot keels). The utilisation of the prosthesis components (heel and fore-foot keels) with different stiffness properties would have an effect on the assessed variables, predominantly on the ankle joint power. Nevertheless, the heel and fore-foot keel components were used intra-subject variability and were specified by the same experienced prosthetist according to the weight, height and gait of the patient to ensure safe, comfortable and efficient locomotion.

Prior to data collection amputee participants had to be familiarised with each foot type (*Elan, Epirus*) by walking on the level floor of the laboratory for approximately 20 minutes, so amputees had limited accommodation time. In addition, data for each attachment condition for each participant was collected in a single session. The results might be different if the accommodation time was longer. However, all participants used or had used one habitually prosthetic attachment with a hydraulic articulated 'ankle'. To eliminate 'order effects', the order of prosthetic feet (*Epirus* and *Elan*) was counterbalanced across participants. To eliminate 'bias effects' with the *Elan* ankle-foot, active and non-active modes were 'blind' for amputees in randomly counterbalanced order.

Ramp-descent in amputees was undertaken with them using a prosthetic foot with the same heel and fore-foot keels but with three different ankle articulation mechanisms (MC-AF, nonMC-AF, elasticAF). Thus, a limitation was that there was no comparison with a non-articulating prosthetic ankle-foot. This experimental parameter did not include a fourth foot condition and an increased number of trials would have been problematic due to potential fatigue issues for participants, and/or that the biomechanical compensation required when using such feet might have 'carry-over' effects of the other foot conditions. The use of the non-articulated prosthetic device would be particularly problematic as all amputee participants habitually used an articulating ankle-foot device (see participant details chapter three).

In experimental chapters four and five, able-bodied participants utilised a custom made ankle-foot-orthosis (AFO) which had some design limitations. The length of AFO's steel struts that were located either side of the shank could only be adjusted in steps of 10 mm. Nevertheless, this would permit, no more than 5mm misalignment in vertical placement along the shank when adjusting the strut to ensure alignment with the ankle-knee axis and accurate positioning of the gastronimius pads. This alignment would not affect the study results or harm participants. The AFO restricted ankle movement in the sagittal plane to around ±3-5 degree of plantar/dorsi-flexion it did not eliminate it. This small amount of movement was due to the AFO flexing and/or relative motion between the AFO and the soft tissues of the shank. However, the amount of flexing/relative motion was likely to have varied across participants depending on anthropometry, shoe size etc., and this inter-subject variability was considered to have negated any systematic effects.

The use of kinetic parameters has some nuisances that should be acknowledged. In the biomechanical model, the segments used were considered as rigid. The effect of friction in joints was not counted because of negligible friction forces. Although, in patients with joint disease (i.e. osteoarthritis) the modelling of joint friction forces should be considered (Zajac and Winters 1990). The research of Drewniak and colleagues has examined and suggested prediction of joint friction, lubrication, and wear (tribology) by using different methods to assess the frictional properties of articular cartilage (Drewniak et al. 2009). The ramp has only one force platform structure integrated into the construction. Hence, kinetic data were recorded at a single trial so it only the involved limb that landed on the force platform. The limited number of force plates is the result of the complex installation required and the associated high cost. Force plates have to be installed in a laboratory which commonly have a short walk away, so participants have to adjust step length and walking speed according to the environment (Wearing et al. 2001). However, the study of Astephen Wilson demonstrated that the biomechanics of gait adapts according to the surrounding environment (Astephen Wilson 2012). Additionally, a number of researchers have examined the effects of different prosthetic attachment, bracing and/or limb impairment on the contralateral (un-involved) limb and the symmetry between limbs during overground gait (Nolan and Lees 2000; Nolan et al. 2003; Franz et al. 2012; De Asha et al. 2013b). Use of a prosthetic device in TTs has an impact on the intact/non-effected limb and may lead to developing osteoarthritis and/or lower back pain (Hurley et al. 1990; Kulkarni et al. 1998; Gailey et al. 2008). Recording data from the non-involved limb landing was not included in the current thesis and thus should be examined in future work.

Investigations of amputee gait often involve modelling of the prosthetic foot and ankle. In experimental chapter seven of the thesis, the mechanical power absorbed and returned by the ankle-foot prosthetic device was examined. To examine power absorption/return an energy flow technique was employed (Takahashi et al. 2012). This approach models the foot-ankle device as a unified deformable segment (*UDS*). The advantage of this model over inverse dynamics which uses rigid segments and mechanical joints is that there is no requirement to define a prosthetic 'ankle' joint centre. This modelling approach is limited because analysis in the *UDS* cannot distinguish how energy transferred within the structure of prosthesis. Thus, it would not be possible to examine certain prosthetic foot components or even distinguish it from the footwear due to *UDS* modelling. Another limitation of *UDS* is that the model delivers only scalar quantity which could be examined only during the stance phase due to the requirements of GRFs. Finally, as an anatomically relevant model, the *UDS* power could be affected by soft tissue movement (Manal et al. 2002). Future studies of lower-limb amputees should employ a prosthetic foot model that includes the specific functionality of devices. The modelling of prosthetic devices could include the movement between amputees' residual limb and prosthetic socket due to the effects of friction on the load transfer. Previous research stated this problem to evaluate the effects of friction during load (Zhang et al. 1996). Soft tissues of the residuum inside a prosthetic socket are positioned under specific conditions. The following conditions could have an effect on the residuum-socket interface. Initially, the fitting of the socket provides the amount of load that soft tissue takes during the locomotion cycle. The friction between the socket and soft tissue which could be affected by high humidity inside the socket where the tissue has accumulated sweat. Lastly, soft tissue could also have a (allergic) reaction on the socket or interface materials (Mak et al. 2001). Although the stress on residual limb socket interface can be measured, an accurate modelling of the load transfer remains challenging due to the complexity of the measurement and the absence of a consistent system (Pirouzi et al. 2014).

Further research on ramp descent might explore different populations, such as trans-femoral amputees, older adults, and other patients with lower-limb impairments that will be beneficial in understanding the abilities and/or compensations in these groups. Such an understanding could lead to better treatment programs, rehabilitation training, and prosthetic device design. The objective of rehabilitation training in lower-limb amputees is to maximise mobility with the prosthesis (Taylor et al. 2008) as well as improve muscle strength of the remaining joints (Isakov et al. 1996a). The use of an inclined surface in rehabilitation might be employed to reach those objectives (Vrieling et al. 2008). For example, the research of Yano et al. (2015) shows how the use of the ramp rehabilitation program exercises the vastus medialis muscles with hip flexion/extension in order to lessen the risk of falling (Yano et al. 2015). The examination of different patient populations during ramp descent may suggest further insights of body biomechanics on inclined surfaces with further suggestions for revisions in the rehabilitation process to improve the patients' safety.

Further work is also required to measure energy cost during ramp descent. There was no investigation of the metabolic cost in this thesis. However, future research should examine the effect of MC-AF on ramp descent. Ramp descent compared to overground gait involves an increase of lead limb knee work in able-bodied individuals (Lay et al. 2007) (Chapter five) and TTs (Vickers et al. 2008) (Chapter seven).

The thesis has acknowledged the impact of the prosthetic ankle-foot devices and restricted ankle function on a stance phase during ramp descent. Despite this, future research could investigate performance of the contralateral limb during the swing phase, according to the functionality of the prosthetic 'ankle' during prosthetic limb stance. For example, examining the minimum toe clearance during ramp descent is critical, as increased potential gravitational energy would increase the repercussion of a fall. Overground gait has minimum toe clearance during mid-swing (Murray et al. 1966; Winter 1992), so the compensations that occur during contralateral single-limb-support could have an effect on safe toe clearance. There have been several studies in the literature reporting that prosthetic 'ankle' functionality did not affect minimum toe clearance on the prosthetic and intact limb during overground gait (Wurdeman et al. 2012; Johnson et al. 2014; De Asha and Buckley 2015). However, whether such 'ankle' function would affect minimum toe clearance during ramp descent is unknown. This examination of minimum toe clearance would be beneficial to the rehabilitation program.

8.3 Final Conclusions

The current thesis contributes an important additional knowledge in biomechanical adaptations according to prosthetic ankle-foot articulation during ramp descent that occurs in TTs. Currently, there is no scientific literature of biomechanical adaptations that specifically investigated different prosthetic ankle-foot articulations during ramp descent in active, trans-tibial amputee population except published literature that has stemmed from the thesis.

The current thesis has demonstrated that unilateral ankle restriction in able-bodied individuals and unilateral TTs with different articulated ankle-foot devices have comparable biomechanical adaptations during ramp descent. Furthermore, this thesis suggested that the use of a microprocessor-controlled hydraulically articulated a with incorporating a dynamic response foot device (adaptive) has biomechanical benefits and could lessen the metabolic cost during the ramp descent for active, unilateral TTs. Findings supported that the use of such articulation reduces knee joint compensation during the ramp descent. Although, the ‘ankle’ articulations did not have an effect on the symmetry between limbs or impact on the contralateral side.

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Appendices

Appendix 1

LOCOMOTOR CAPABILITIES INDEX IN AMPUTEES (LCI)
Whether or not you wear your prosthesis, at the present time, would you say that you are "able" to do the following activities with your prosthesis on?
Please circle the number that best describes your capability.

ITEM	NO	YES, if someone helps me	YES, if someone is near me	YES, alone, with ambulation aids	YES, alone, without ambulation aids
1. Get up from a chair	0	1	2	3	4
2. Walk indoors	0	1	2	3	4
3. Walk outside on even ground	0	1	2	3	4
4. Go up the stairs with a handrail	0	1	2	3	4
5. Go down the stairs with a handrail	0	1	2	3	4
6. Step up a kerb	0	1	2	3	4
7. Step down a kerb	0	1	2	3	4
Basic Activities Score					
1. Pick up an object from the floor (when you are standing up with your prosthesis)	0	1	2	3	4
2. Get up from the floor (e.g. if you fall)	0	1	2	3	4
3. Walk outside on uneven ground (e.g. grass, gravel, slope)	0	1	2	3	4
4. Walk outside in bad weather (e.g. snow, rain, ice)	0	1	2	3	4
5. Go up a few steps (stairs) without a handrail	0	1	2	3	4
6. Go down a few steps (stairs) without a handrail	0	1	2	3	4
7. Walk while carrying an object.	0	1	2	3	4
Advanced Activities Score					
Total Score					

Appendix2

PARTICIPANT CONSENT FORM

Importance of ankle motion to gait on slopes
Researcher – Mr Vasily Struchkov, University of Bradford School of Engineering, Design and Technology

Initial

1. I confirm that I have read and understand the information provided for the above study. ☐
2. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily ☐
3. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving a reason and that this will not affect my medical care or legal rights. ☐
4. I understand that any personal information collected during the study will be anonymised and remain confidential. ☐
5. I agree to take part in the above study. ☐

Name of Participant Date
Signature

Name of Researcher Date
Signature

Name of Person taking consent Date
Signature
(if different from researcher)

Note: When completed, 1 copy for participant and 1 copy for researcher

Appendix 3

PARTICIPANT CONSENT FORM

Effects of adaptive hydraulic ankle damping on ramp and overground gait in unilateral trans-tibial amputees.

Researcher – Mr Vasily Struchkov, University of Bradford School of Engineering, Design and Technology

	Initial
1. I confirm that I have read and understand the information provided for the above study.	☐
2. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily	☐
3. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving a reason and that this will not affect my medical care or legal rights.	☐
4. I understand that any personal information collected during the study will be anonymised and remain confidential.	☐
5. I agree to take part in the above study.	☐

Name of Participant	Date	Signature
Name of Researcher	Date	Signature
Name of Person taking consent *(if different from researcher)*	Date	Signature

Note: When completed, 1 copy for participant and 1 copy for researcher

Appendix 4

Participant code .... Participant initials.......... Date............

BASELINE DATA PROFORMA

Effects of adaptive hydraulic ankle damping on ramp and overground gait in unilateral trans-tibial amputees.

Are you feeling well today Y / N

Consent form signed Y / N

Meets inclusion criteria Y / N

DoB ……………………Gender M/ F

Amputation Side L / R Time since amputation ……..

Cause ……………………………………………

How long had current prosthesis …………… Hours per day typically worn ……………

Any problems with residuum Y / N, if Y record exact problems ………………………………..

……………..

Relevant medical conditions …………………………………..Medications………………………

Phantom limb symptoms…………Y/N or N/A Phantom pain ……………. Y/N or N/A

Socket ……………………Socket length…………

Prost Foot ………………………Stiffness ………Size ………Build height (170/175mm/…….)

Liner ……………………………………..

Weight…… Height ….. Foot Length ….. Shank pylon Length ……….Alignment………...

Intact foot length ……………..

Toe	Vertical	L	R
	Horizontal	L	R
Heel	Vertical	L	R

Mass of a shoe ……………………..g Mass of prosthesis………….g

Residuum Length Circumference Prox Dist

Microprocessor settings……………………………………………………………….

Habitual visual correction worn when walking:

*none *contact lenses *spectacles (varifocal/bifocal/single distance vision) Years……

Visual acuity L R Bin Dominant eye L / R

LCI score……………………

Appendix 5

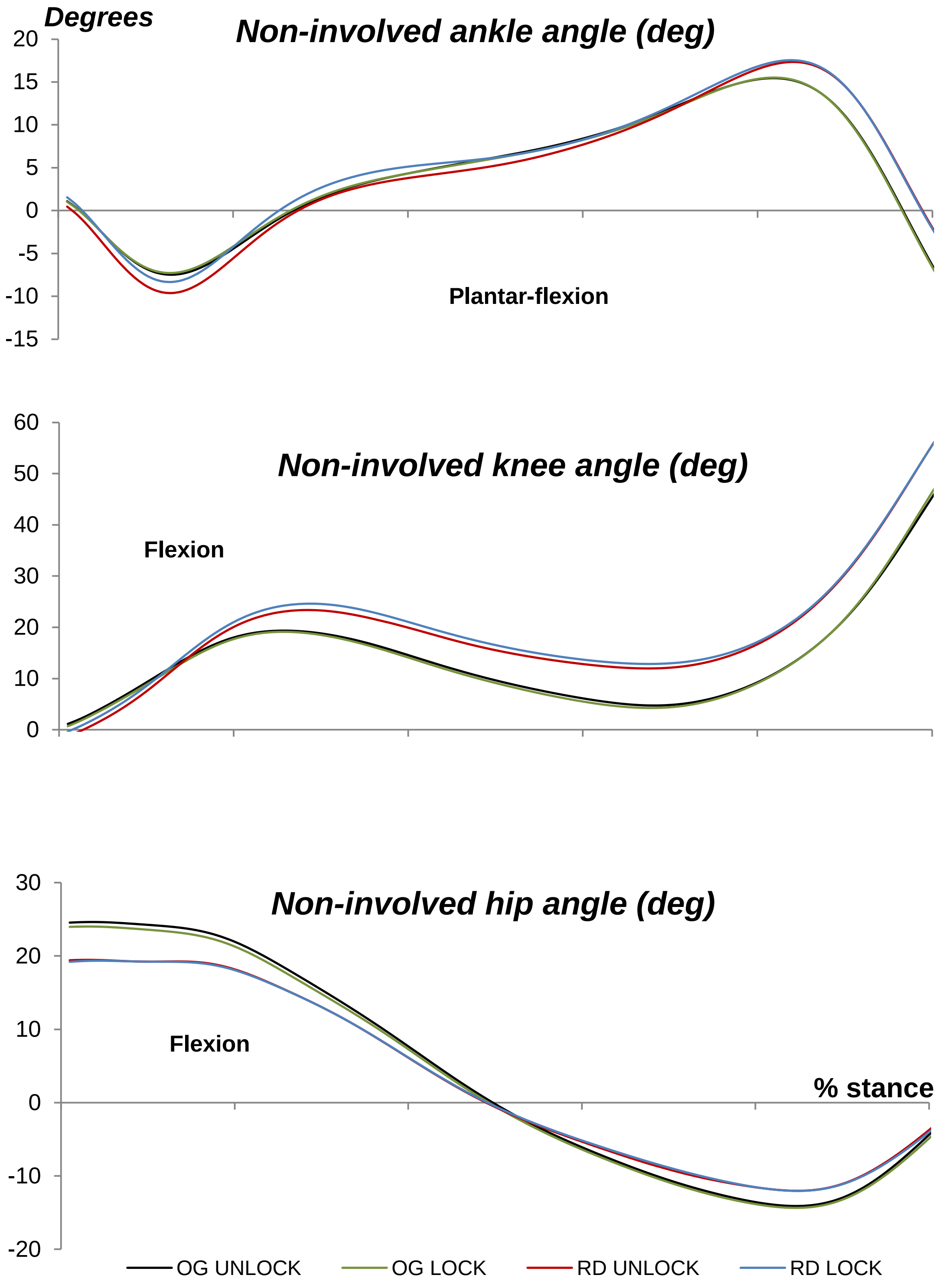
The force structure of the inclined block was built and assessed over 5 trials in Visual 3D software (C-Motion Inc., Germantown, MD).

	CT Orent Error mean	CT Orent Error SD	CT Tip Diff mean	CT Tip Diff mean	CT Tip Diff mean	CT Tip Diff SD	CT Tip Diff SD	CT Tip Diff SD
			X	Y	Z	X	Y	Z
1	0.768	1.437502	-0.01274	-0.01256	-0.00008	0.00215	0.002726	0.00275
2	0.911	1.756388	0.009311	-0.01513	-0.003195	0.0028	0.003789	0.000318
3	0.722	1.036342	0.007297	0.012984	-0.00797	0.001386	0.003738	0.000318
4	1.04	2.963979	-0.01628	0.014111	0.001586	0.005905	0.005045	0.000468
5	0.782	0.921827	-0.00447	-0.00128	-0.000092	0.001576	0.001576	0.000204
mean	0.8	1.6	-0.003	0.000	-0.002	0.003	0.003	0.001
SD	0.1	0.8	0.012	0.014	0.004	0.002	0.001	0.001

Appendix 6

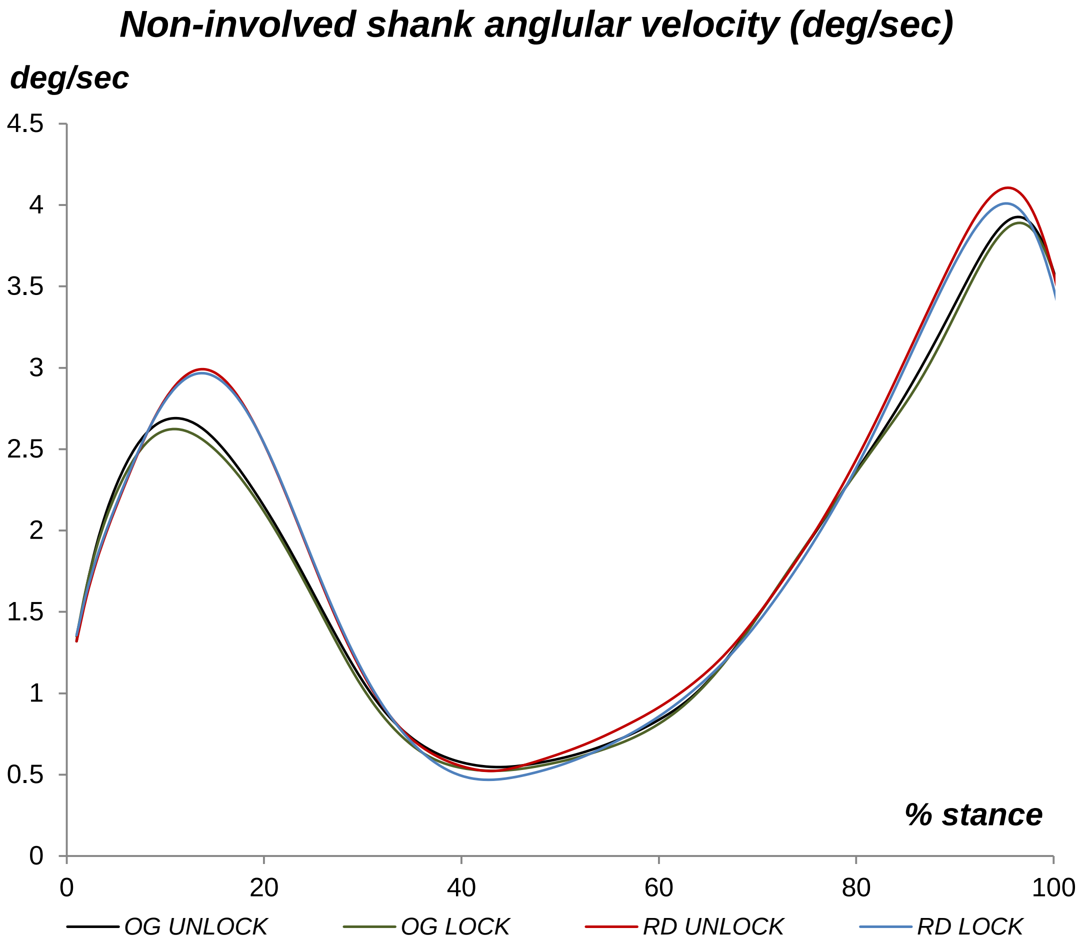
Number	Position of a marker	Labels
1	Headband: Anterior left	ANT_HEAD_L
2	Headband: Anterior right	ANT_HEAD_R
3	Headband: Posterior left	POST_HEAD_L
4	Headband: Posterior right	POST_HEAD_R
5	Left acromion process	ACROM_L
6	Right acromion process	ACROM_R
7	Jugular notch	STURNUM
8	Xiphoid process	XIP_PROC
9	C7 vertebrae	C7
10	T8 vertebra on spine	T8
11	Sacrum cluster: Superior	SACR_ANT
12	Sacrum cluster: Left	SACR_L
13	Sacrum cluster: Right	SACR_R
14	Sacrum cluster: Inferior	SACR_INF
15	Left iliac crest	ILCREST_L
16	Right iliac crest	ILCREST_R
17	Left great trochanter	GTROC_L
18	Right great trochanter	GTROC_R
19	Left thigh plate: Proximal anterior	UL_PR_ANT_L
20	Left thigh plate: Proximal posterior	UL_PR_POST_L
21	Left thigh plate: Distal anterior	UL_DI_ANT_L
22	Left thigh plate: Distal posterior	UL_DI_POST_L
23	Left knee: Medial femoral epicondyle	KNEE_MED_L
24	Left knee: Lateral femoral epicondyle	KNEE_LAT_L
25	Left shank plate: Proximal anterior	LL_PR_ANT_L
26	Left shank plate: Distal anterior	LL_DI_ANT_L
27	Left shank plate: Proximal posterior	LL_PR_POST_L
28	Left shank plate: Distal posterior	LL_DI_POST_L
29	Left foot: Medial malleolus	MAL_MED_L
30	Left foot: Lateral malleolus	MAL_LAT_L
31	Left foot: Metatarsal head 1	MTH1_L
32	Left foot: Metatarsal head 5	MTH5_L
33	Left foot: Anterior edge	TOE_L
34	Left foot: Midfoot medial edge	FOOT_MED_L
35	Left foot: Midfoot lateral edge	FOOT_LAT_L
36	Left foot: Heel	HEEL_L
37	Right thigh plate: Proximal anterior	UL_PR_ANT_R
38	Right thigh plate: Proximal posterior	UL_PR_POST_R
39	Right thigh plate: Distal anterior	UL_DI_ANT_R
40	Right thigh plate: Distal posterior	UL_DI_POST_R
41	Right knee: Medial femoral epicondyle	KNEE_MED_R
42	Right knee: Lateral femoral epicondyle	KNEE_LAT_R
43	Right shank plate: Proximal anterior	LL_PR_ANT_R
44	Right shank plate: Proximal posterior	LL_PR_POST_R
45	Right shank plate: Distal anterior	LL_DI_ANT_R
46	Right shank plate: Distal posterior	LL_DI_POST_R
47	Right foot: Medial malleolus	MAL_MED_R
48	Right foot: Lateral malleolus	MAL_LAT_R
49	Right foot: Metatarsal head 1	MTH1_R
50	Right foot: Metatarsal head 5	MTH5_R
51	Right foot: Anterior edge	TOE_R
52	Right foot: Midfoot medial edge	FOOT_MED_R
53	Right foot: Midfoot lateral edge	FOOT_LAT_R
54	Right foot: Heel	HEEL_R

Appendix 7



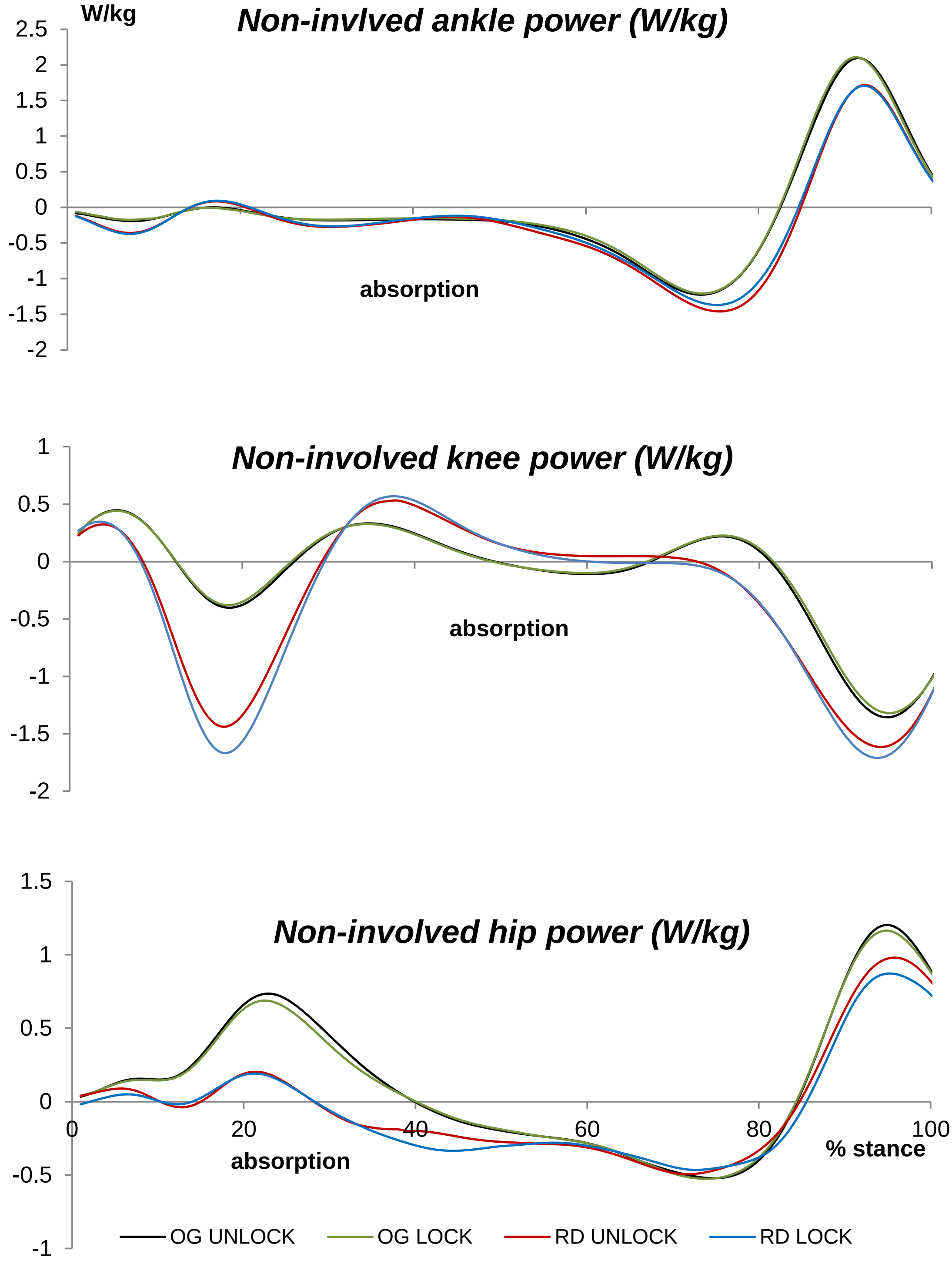
Non-involve limb (ankle, knee, hip) joints angular displacement (deg.) normalised to 100 points (stance phase), averaged across 20 participants. Positive angles are plantar-flexion for the ankle and flexion of the knee and hip joints. (OG UNLOCK – overground non-restricted; OG LOCK –overground restricted; RD UNLOCK –ramp descent non-restricted; RD LOCK –ramp descent restricted).

Appendix 8



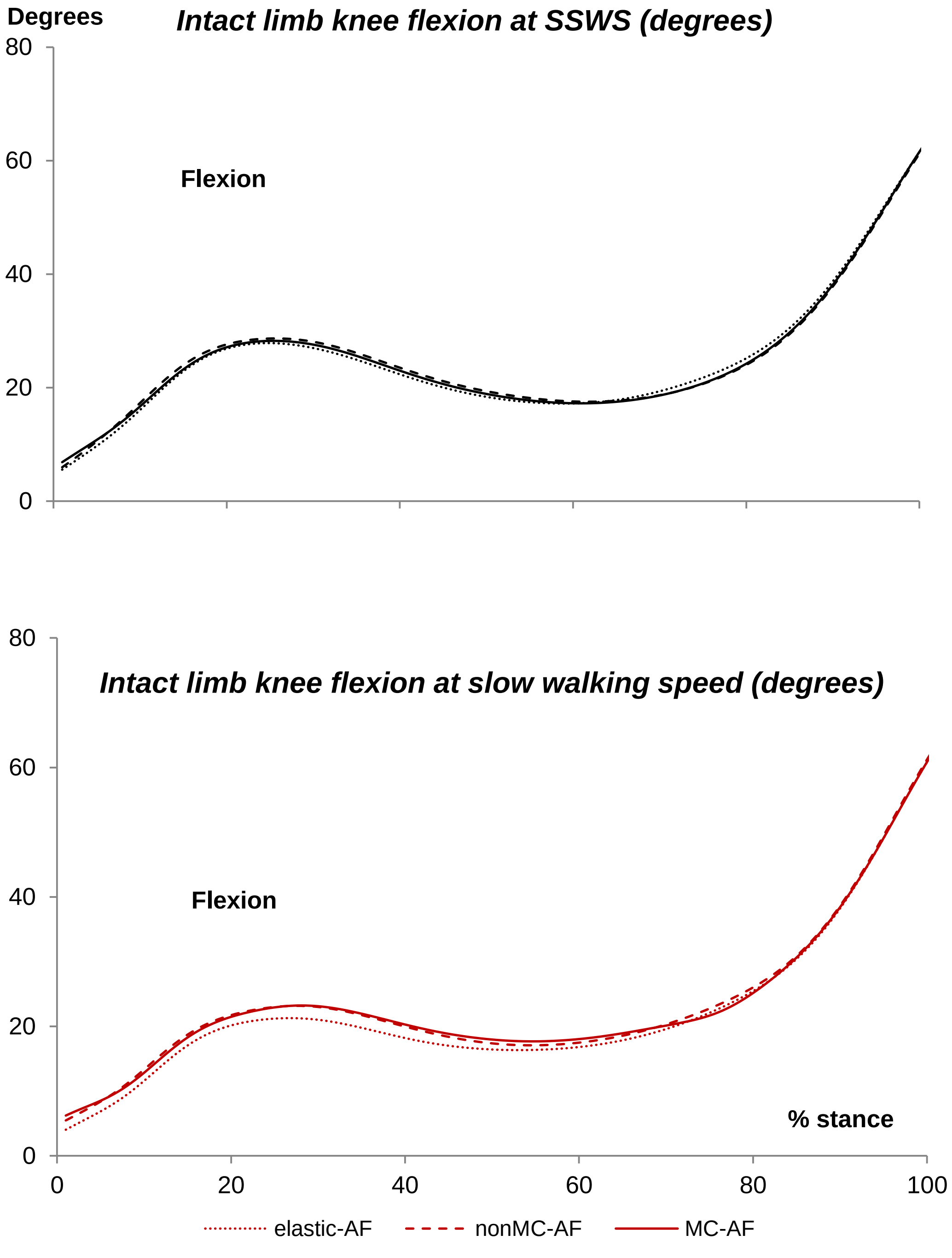
Mean Shank angular velocity normalised to 100 points (stance phase), and ensemble averaged across 20 subjects. (OG UNLOCK –overground non-restricted; OG LOCK –overground restricted; RD UNLOCK –ramp descent non-restricted; RD LOCK –ramp descent restricted).

Appendix 9



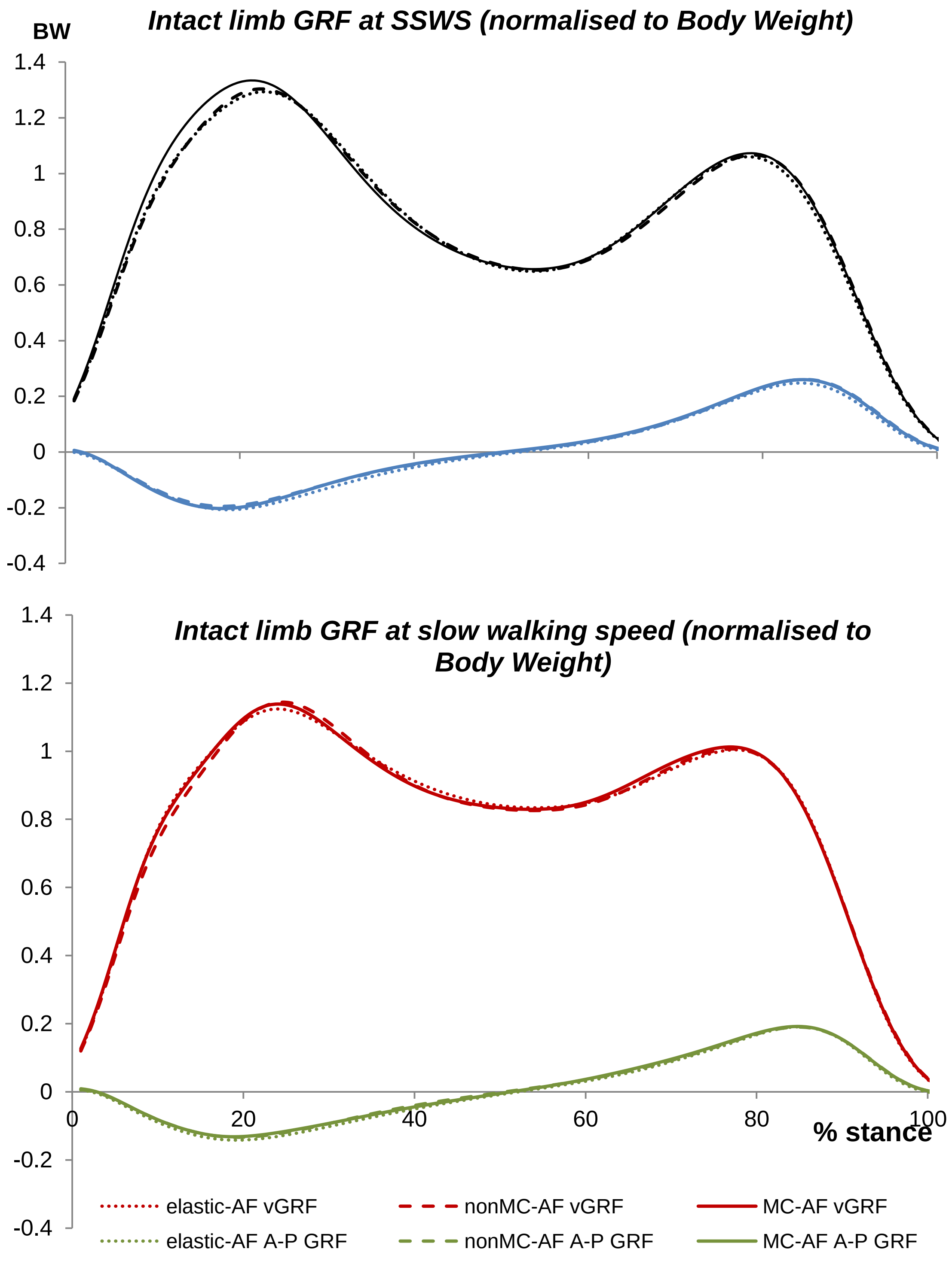
Mean of Non-involve (left) limb a/ ankle joint power (W/kg); b/ knee joint power (W/kg); c/ hip joint power (W/kg); d/ limb total rotational power (W/kg) normalised to 100 points (stance phase), and ensemble averaged across 20 subjects (ankle, knee, hip) joints angular displacement (deg.) normalised to 100 points (stance phase), averaged across 20 participants. (OG UNLOCK –overground non-restricted; OG LOCK –overground restricted; RD UNLOCK –ramp descent non-restricted; RD LOCK –ramp descent restricted).

Appendix 10



Intact limb knee joint angular displacement (deg.) normalised to 100 points (stance phase), averaged across 9 TT participants. Positive angles flexion for the knee joint for (black line) self-selected walking speed (SSWS) and (red line) slow walking speed when using the elastic-AF (dotted line), nonMC-AF (dashed line), MC-AF (solid line) ankle-foot.

Appendix 11



Intact limb Vertical ground reaction forces (vGRF) and anterior-posterior ground reaction forces (A-P GRF) (normalised to Body Weight) normalised to 100 points (stance phase), averaged across 9 TT participants. Self-selected walking speed (SSWS) vertical ground reaction forces (vGRF) (black) and anterior-posterior ground reaction forces (A-P GRF) (blue); slow walking speed vGRF (red) and A-P GRF (green) when using the elastic-AF (dotted line), nonMC-AF (dashed line), MC-AF (solid line) ankle-foot.